Department of Veterans Affairs

Journal of
Rehabilitation Research and Development

November 1994, Vol. 31, No. 4

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Veterans Health Administration
Rehabilitation Research and Development Service

CALL FOR PAPERS

We would welcome submission of manuscripts in the fields of Prosthetics and Orthotics; Spinal Cord Injury and Related Neurological Disorders; Communication, Sensory and Cognitive Aids; and Gerontology. Guidelines for submission of manuscripts may be found on page ii.

Editor
Tamara T. Sowell

LETTERS TO THE EDITOR SECTION

Interested readers are encouraged to engage in an exchange of information through this Section. Letters should relate specifically to material published in the Journal of Rehabilitation Research and Development. We request that the following information be supplied: full name of the author of the article, title of the article, Volume and Issue number, the page number on which the article appeared. In addition, we request that the letter contain the full name and academic degree of the letter writer, along with the facility that the writer represents.
We hope to open up an ongoing dialogue between professionals as a means of exchanging information and sharing diverse opinions among disciplines.

Editor
Tamara T. Sowell

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Journal of Rehabilitation Research and Development
NOTICE TO CONTRIBUTORS

Purpose and Scope

The *Journal of Rehabilitation Research and Development*, published quarterly, is a scientific rehabilitation research and development publication in the multidisciplinary field of disability rehabilitation. General priority areas are: Prosthetics and Orthotics; Spinal Cord Injury and Related Neurological Disorders; Communication, Sensory and Cognitive Aids; and Gerontology. The *Journal* receives submissions from sources within the United States and throughout the world.

Only original Scientific Rehabilitation Research and Development papers (including Preliminary Studies) will be accepted.

Technical Notes describing techniques, procedures, or findings of original scientific research may be submitted. Clinical Reports are of particular interest. These may be reports of an evaluation of a particular prototype developed, a new clinical technique, or any other topic of clinical interest. Letters to the Editor are encouraged. Books for Review may be sent by authors or publishers. The Editor will select reviewers.

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EDITOR'S NOTE

To Peer Reviewers

In this, the final issue of 1994, I would like to take the opportunity to thank the entire editorial board and all of our peer reviewers for their thoroughness, tireless assistance, and forbearance in reviewing manuscripts submitted to the *Journal*.

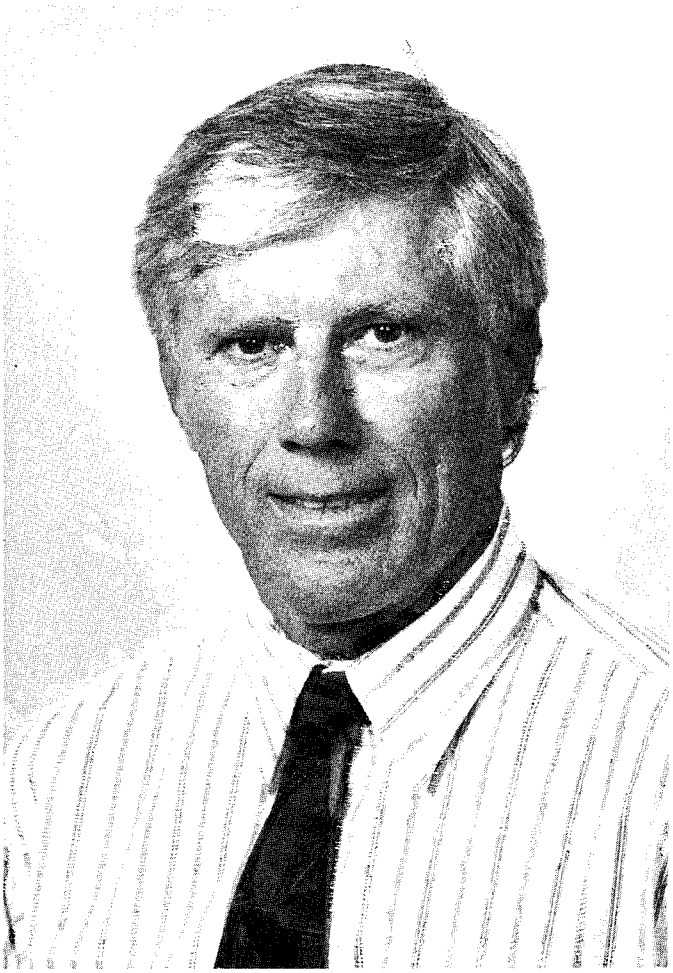
I have often received letters from authors expressing their appreciation for the guidance received from the comments of the anonymous peer reviewers. Authors stated that the information helped them considerably in the revision of their papers. On behalf of the authors, I thank you again.

In the coming year, I hope I may again count on you, the experts, to provide your usual superior peer review for our manuscripts and thereby be assured that the articles accepted for publication in the *Journal* will continue to enrich the body of scientific literature.

Tamara T. Sowell

GUEST EDITORIAL

From Curing to Caring



Donald R. McNeal, Ph.D.
Director, Rehabilitation Engineering Program,
Rancho Los Amigos Medical Center, Downey, CA

I recently read a book written by Daniel Callahan entitled "What Kind of Life: The Limits of Medical Progress." Some of Callahan's ideas are controversial, but whether you agree with them or not, they are worth considering as we debate the future course of healthcare in the United States. He also has some thoughts about federal expenditures for research and development that I shall get to later. With apologies to Callahan for oversimplification of his ideas and for any unintended distortion of his basic thesis, let me try to summarize some of his major points.

Callahan begins with the obvious fact that healthcare costs in the United States are rising rapidly despite intensive efforts beginning in the 1970s to control costs. National healthcare expenditures have risen from 5.9 percent of our gross national product (GNP) in 1965 to 11.2 percent in 1987. We already spend a higher percentage of our GNP on healthcare than any other industrialized nation, and without major reform, the percentage is expected to increase to 15 percent by the year 2000.

Three major factors that contribute to rising healthcare costs are cited. The first is the increase in the elderly population, the group that requires higher expenditures for healthcare than any other age group. The number of people over the age of 65 (approximately 30 million) is expected to double within the next 30 years. The fastest-growing age group consists of those over the age of 85, and that number could well triple over the same period. The second factor is the explosion of technological advances that have improved medical care and extended life. These technologies are expensive, and they often extend life at the cost of other complications and additional major expenses. Finally, there is the public demand that all individual needs for cure be met. The healthcare system that we have created thus seeks to conquer all disease and to prolong life at all cost. Callahan contends that this is a battle that we cannot win, but one that will consume funds without limits.

These three factors—an aging society, endless technological advances, public demand that all disease be cured—will continue to drive healthcare expenditures up, unless there are fundamental changes in our values and goals for our healthcare system. Callahan argues that there must be a shift in emphasis from curing to caring. He contends that everyone should have a minimally adequate level of caring, but people should not expect society to meet all individual needs for cure.

The great improvements in health worldwide have come about in three stages. The first, which took place from the 17th through the early 20th centuries, produced better nutrition, sanitation, and general living conditions. The second occurred during the late 19th to the mid 20th centuries, when we achieved the conquest of most infectious diseases through vaccinations and antibiotics. From the mid 20th century on, we have seen the major technological advances that have resulted in improved surgical

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techniques, intensive care units, improved rehabilitation, and organ transplants.

Callahan believes that if society did nothing more than keep the conditions of the first two stages in good working order, it would ensure long and healthy lives for the majority of the population. But he contends that we can and should do more. He puts forward the following goals for our healthcare system:

> "The primary goal of the healthcare system should be to provide those general measures of public health and basic medical care most likely to benefit the common health of the population as a whole, and to ensure that every person in the society receives care, comfort and support in the face of illness, aging, decline and death. The secondary goal of the system should be—within the limits of a reasonable level of healthcare expenditures in relationship to other societal needs—to pursue a basic understanding of the causes of illness and death, and to aspire to the cure of those illnesses that bring premature death and thwart common human aspirations."

Whether you accept Callahan's vision of a national healthcare system, it seems clear that we are moving toward a system that will provide a basic level of healthcare for all (universal coverage) with limitations on the amount of coverage for most Americans. Most likely, these limits will apply to many of the more expensive treatments or therapeutic modalities, especially those that cannot demonstrate clear, long-term beneficial outcomes.

Given this direction for our healthcare system, it is imperative that we review the priorities for committing limited funds for research and development. Clearly, we cannot afford to invest limited tax dollars in the development of treatment programs or equipment which provide little or no additional benefit over existing techniques. The more difficult question is whether federal funds should be invested in R&D projects which may lead to treatments that are efficacious, but are also extremely costly—treatments which our healthcare system may not support.

Callahan says we should not. He believes that we should not develop new technologies for saving lives until we can meet the needs of those who have already survived and whose lives promise long suffering, whether physical or psychological. He proposes that a technology should be judged by its likelihood of enhancing a good balance between the extension and saving of life and the quality of life. It should foster the rounded well-being of persons, not simply one-dimensional improvements that benefit some aspect of individual well-being at the expense of others.

In his book, Callahan does not specifically address rehabilitation or rehabilitation research. I suspect, however, that he would strongly support research for rehabilitation because he states that "research priorities should be directed to improving the quality of life of those already burdened with illness or disability, rather than determining how to further extend life." I also suspect that he would support costly programs to enable the lame to walk or the blind to see only after sufficient funds had been invested in R&D projects to improve the health and well-being of persons with disabilities and to enable them to participate fully in all of life's activities. In any case, we must ensure that research funds are used wisely to support projects that have a high likelihood of success and which will result in outcomes that are cost-beneficial and consistent with national healthcare policies.

Donald R. McNeal, Ph.D.

Clinical Relevance for the Veteran

SUMMARY OF SCIENTIFIC/TECHNICAL PAPERS IN THIS ISSUE

Balance and Stabilization Capability of Paraplegic Wheelchair Athletes.

Pierre Louis Bernard, PhD; Edouard Peruchon, PhD; Jean-Paul Micallef, PhD; Claude Hertogh, PhD; Pierre Rabischong, MD *(p. 287)*

Purpose of the Work. In order to prevent wheelchair athletes from having painful accidents, an investigator would need to know what the individual plans to do and whether that individual is capable of doing it. Performance analysis of the body balance regulation system in the sitting position is a basic measurement for evaluating function. The aim of the present study was to define balance control capabilities of paraplegic wheelchair athletes with different levels of injuries to the spinal cord. **Subjects/Procedures.** Acceleration transmitted to the head was measured to determine the balance performances of the subjects. Moreover, to define the strategies of the wheelchair users, the relative contribution of cervical and thoracic spines to balance was evaluated by measuring acceleration at various sites along the spinal column. Two groups, each with six athletes with paraplegia, one composed of "high paraplegic athletes" (HPA), with a neurological level between T4 and T8, and one composed of "low paraplegic athletes" (LPA), with a neurological level between T11 and L5, were selected. A third group consisting of six "able-bodied healthy athletes," was also used to provide us with a functional reference. **Results.** There seems to be a relationship between the stabilization capability and the neurological level of the subject. On the whole, we observed that the damping factor values at the head decreased with the intensity of stress. The authors attempted to differentiate balance strategies in the LPA and HPA groups through analysis of the relative contributions to damping of the thoracic and cervical spinal segments. The first results show an increasing tendency of neck reflex stiffening according to the neurological level. **Relevance to Veteran Population.** This study reveals the balance capabilities of athletes with paraplegia and offers a means of analyzing their behavior under well-defined mechanical conditions. This method could be a relevant quantitative indicator for assessing the ability of the subject with paraplegia to obtain efficient body balance in the sitting position.

Pierre Louis Bernard, PhD

A Survey of Marginal Wheelchair Users.

Barnaby A. Perks, BSc, MSc; Rosalind Mackintosh, Dip COT; Colin P.U. Stewart, MB, ChB, MD, D Med Rehab; Geoff I. Bardsley, B Eng, PhD, MBES *(p. 297)*

Purpose of the Work. As part of a project entitled "The Determination of Optimum Wheel Configurations for Wheelchair Users," this survey was carried out to identify and describe wheelchair users in Tayside, Scotland, who have limited self-propulsion. **Subjects/Procedures.** From a population of over 3,000 wheelchair users, 83 users were selected for interview. The interview was in the form of a questionnaire designed to highlight factors limiting successful wheelchair propulsion and daily use. **Results.** Survey results indicated that marginal users represent approximately 15 percent of the occupant-propelled wheelchair population. The average age was 48 years and the largest diagnostic category was Multiple Sclerosis. Fifty-nine percent of those questioned felt that their wheelchairs were not adequate for their requirements. **Relevance to Veteran Population.** The survey results suggest direct implications for the assessment, choice of wheelchair, and on-going review of users' needs.

Barnaby A. Perks, BSc, MSc

Distributed Random Electrical Neuromuscular Stimulation: Effects of the Inter-stimulus Interval Statistics on the EMG Spectrum and Frequency Parameters.

Yuan-Ting Zhang, PhD; Philip A. Parker, PhD; W. Herzog, PhD; A. Guimaraes A, PhD *(p. 303)*

Purpose of the Work. One basic measurement technique that is commonly used for evaluating muscular fatigue is the median frequency of the power spectrum of the electromyographic (EMG) signal. The purpose of this study was to demonstrate how different electric neuromuscular stimulation (ENMS) techniques (periodic versus random) affect spectral measurements of the EMG signal, particularly the median frequency. **Procedures.** To mimic the nonperiodic firing of active motor units as observed during voluntary muscle contraction, a random inter-stimulus interval was introduced in the electric neuromuscular stimulation protocol. A mathematical model, similar to that of EMG signal generation, was used to illustrate the dependence of the median frequency on the stimulation rate and on the pulse nonperiodicity as measured by the coefficient

of the variation of the inter-stimulus intervals. Experiments using periodic and random stimulations were performed on the cat soleus muscle. **Results.** Periodic stimulations were found to be limited in reproducing EMG spectra similar to those obtained during voluntary muscle contractions. The dependence of the median frequency on the stimulation rate during periodic stimulations is much stronger than that during random stimulations which mimic the behavior of nonperiodic firing of active motor units. **Relevance to Veteran Population.** Knowledge of relationships between spectral and stimulation measurements may help in the understanding of the mechanism underlying muscle fatigue and may lead to improved designs of electrical neuromuscular stimulators. This applies to functional electrical stimulation of paralyzed muscle to enable a person to walk.
Yuan-Ting Zhang, PhD

Initial Clinical Evaluation of a Wheelchair Ergometer for Diagnostic Exercise Testing: A Technical Note.

W. Edwin Langbein, PhD; Kevin C. Maki, MS; Ming H. Hwang, MD; Pat Sibley, RN; Linda Fehr, MS
(p. 317)

Purpose of the Work. Numerous patients receiving health care from the VA medical system have lower limb disabilities (LLD). Previous research has demonstrated that lower limb disabled possess greater than average risk of acquiring coronary artery disease (CAD). The purpose of this study was to evaluate a new wheelchair ergometer (WCE) and exercise test protocol for detection of CAD in men with LLD. **Subjects/Procedures.** Forty-nine patients (63 ± 9 yr) completed WCE tests. Testing was done in stages with increases in work every three minutes. Metabolic and electrocardiogram measurements were taken. **Results.** Fourteen tests were rated positive, 21 negative, and 14 nondiagnostic for exercise-induced ischemia. In 18 patients who underwent coronary angiography, the predictive value was 100 percent (10/10) for a positive, and 50 percent (2/4) for a negative WCE test result. These findings suggest that WCE is a viable initial diagnostic option for some persons who cannot adequately perform treadmill or cycle exercise. **Relevance to Veteran Population.** Wheelchair exercise testing of veterans with LLD can provide clinically useful information about the presence or absence of CAD and facilitate long-term tracking of their cardiovascular health status.
W. Edwin Langbein, PhD

The Southampton Hand: An Intelligent Myoelectric Prosthesis

Peter J. Kyberd, MSc, PhD and Paul H. Chappell, BSc, PhD, CEng, MIEE
(p. 326)

Purpose of the Work. To improve the functional range of hand prostheses while enhancing the cosmetic appearance of prosthetic hands. **Subjects/Procedures.** Two subjects have used different versions of the Southampton hand. One used the hand in the laboratory for a detailed assessment of the ease by which he could operate the hand. The second used a portable system at home and work. **Results.** A comparison of the Southampton Hand with conventional devices showed it was as good as the best elements of any current device, combining the best functional performance of one with the best cosmetic appearance of another. **Relevance to the Veteran Population.** These tests show that it is possible to construct a prosthetic hand which is both more functional and cosmetic than existing designs, while ensuring that it is easier to operate. This would make the technology accessible to a wider user population.
Peter J. Kyberd, MSc, PhD

Toward Classification of Dysphagic Patients Using Biomechanical Measurements.

Narender P. Reddy, PhD; Rony Thomas, MS; Enrique P. Canilang, MD; Judy Casterline, MA, CCSP
(p. 335)

Purpose of the Work. Dysphagia is the impairment of the swallowing mechanism and presents a major problem in the rehabilitation of stroke patients and others with paralyzing diseases. Dysphagia often leads to aspiration, choking, and even death. Current techniques of the assessment are either qualitative and often based on feeling, or involve videofluorography examination which exposes the patient to radiation. The purpose of this investigation was to develop and evaluate techniques for noninvasive measurement of the swallowing, and to develop and evaluate an expert system to classify the risk for aspiration in the patient using these measurements. **Procedures.** Noninvasive measurement of various parameters such as the tongue thrust, swallow pressure, and throat acceleration were obtained and used to classify the patient. This classification was later compared with the classification made by the clinician. **Results.** The biomechanical measurements correlated with the clinical findings well, and the expert system based on biomechanical measurements classified the patient well. **Relevance to Veteran Population.** The noninvasive biomechanical measurements of the swallowing

can be used for continuing assessment on a daily basis so as to identify the patient at risk for aspiration and prescribe proper feeding protocols.
Narender P. Reddy, PhD

A Voice Output Reader for Displays on VCRs and Other Domestic Products.

D. Gareth Evans, BSc, and Paul Blenkhorn, BSc *(p. 345)*

Purpose of the Work. To investigate a non-intrusive method, using a hand-held video camera and image processing, to enable blind people to access the electronic displays on domestic products such as microwave ovens, video recorders, hi-fi systems, etc. **Subjects.** None. **Procedures.** To demonstrate the feasibility of such a system, a prototype has been evaluated by using emulations of a complex Video Cassette Recorder interface with Gas Plasma and Liquid Crystal Displays. **Results.** The system was found to be robust and virtually error free when precise camera alignment was maintained, except for an occasional error in interpreting '0' as '8' on a seven segment display. **Relevance to Veteran Population.** This work can contribute to the production of a 'Display Reader' which would enable blind veterans to operate current and future domestic products more effectively.
D. Gareth Evans, BSc

Department of Veterans Affairs

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Pages 287–296

Balance and stabilization capability of paraplegic wheelchair athletes

Pierre-Louis Bernard, PhD; Edouard Peruchon, PhD; Jean-Paul Micallef, PhD; Claude Hertog, PhD; Pierre Rabischong, Professor, MD

Unité 103 INSERM, "Appareil Moteur et Handicap," 34100 Montpellier, France; UFRAPS Laboratoire, "Sport Santé et Développement," 34080 Montpellier, France.

Abstract—The orientation of paraplegic athletes toward adapted sport activities requires good knowledge of their functional characteristics. Wheelchair locomotion, especially for highly dynamic situations, poses the problem of trunk equilibrium management and head stabilization. The study aimed at designing a quantitative method to assess the ability of paraplegics to obtain trunk balance under dynamic stresses, and to analyze the various balance strategies, according to the spinal lesion level of the subjects. High (HPA) and low (LPA) paraplegic athletes were subjected to four series of antero-posterior stresses of increasing intensity, generated by an oscillating platform. By means of a computerized video-based movement analyzer, acceleration in the saggittal plane was measured at four different spinal levels and, for each one, a damping factor was determined. This factor, computed at the head level, appeared to be representative of the subjects' ability to maintain balance. We attempted to differentiate balance strategies in the LPA and HPA groups through analysis of the relative contributions to damping of the thoracic and cervical spinal segments. The first results show an increasing tendency of neck reflex stiffening according to the neurological level.

Key words: *athletes, balance, damping, paraplegic, postural, wheelchair.*

INTRODUCTION

High performance in wheelchair locomotion requires a combination of low energy cost and optimal comfort. Handrim wheelchair propulsion is a means of locomotion with quite a high exertion demand. Weak propulsive output is obtained (1–4) with relatively high cardiorespiratory stress (5–7). Thorough studies of metabolic and physiological responses to muscular exercise (8–10) and of propulsion techniques (11–13), as well as investigations on wheelchair designs (14–16), have all contributed to reaching the optimal "man-machine" interaction objective.

Due to the increasing interest in wheelchair athletics, it is now critical to improve the man-machine interface, which involves adapting available standard equipment for potential athletic use. This approach takes the functional behavior of wheelchair athletes into consideration and emphasizes prevention of injuries related to handrim wheelchair propulsion (17,18). Head movement and mobilization of the cervical and thoracic spine play major roles during propulsion.

For wheelchair athletes, these spinal areas are crucial for efficient ambulation and trunk balance, thus increasing the risk of musculoskeletal trauma. Various authors have described spine traumatisms and injuries related to wheelchair propulsion and disabilities (19–22), others have carried out kinematic analyses of wheelchair locomotion and related spine movements (23–25). All of these studies have enhanced the overall understanding of spinal behavior in wheelchair users.

Prevention of traumatic risks for wheelchair athletes implies definition of suitably adapted physical activities, which requires precise knowledge of individual real functional potential. Our work falls within the scope of the functional characterization of wheelchair athletes (26,27). Performance analysis of the body balance regulation sys-

Address all correspondence and requests for reprints to: Pierre-Louis Bernard, PhD, Unité 103 INSERM, 395, Avenue des Moulins, 34100 Montpellier, France.

tem in the sitting position is a basic parameter for functional evaluations. The aim of the present study was thus to define balance control capabilities of paraplegic wheelchair athletes with different levels of neurological lesion.

METHODS

Principles

In order to study characteristics of the balance system, wheelchair athletes with paraplegia were subjected to acceleration in the sagittal plane. This acceleration component represents a major traumatic risk for the spine.

Identification of a control system generally consists of analyzing the response of that system to well defined input signals, mainly in the form of square, triangular, or sinusoidal waves. The first one leads to excessive levels of acceleration (shock) and is thus unsuitable for application to persons with paraplegia. In addition, it is technically difficult to implement. Triangular input, which produces (only at the slope inversion) short and powerful acceleration changes, has also been rejected.

Sinusoidal input therefore appeared to be most appropriate in our experimental context. It generates repetitive, alternate, and continuously variable acceleration with time, which eliminates the risk of abrupt stress at controlled levels of frequency and amplitude. Moreover, this type of input has some similarities to real cyclic movements of the trunk of the wheelchair user.

Acceleration transmitted to the head was measured to determine the subjects' balance performances. To define the strategies of the wheelchair users, the relative contribution to balance of cervical and thoracic spines was also evaluated by measuring acceleration at various sites along the spinal column.

Equipment

The study required an original device to move the wheelchair and a computerized video-based movement analyzer for kinematic analysis. Technical difficulties to implement horizontal translational alternate displacements of the wheelchair led us to design the simple moving platform shown in **Figure 1**. This platform oscillates on a horizontal axis (A). It is activated by a servo-controlled hydraulic jack (B) linked to a hydraulic power supply (C). Control is maintained with a function generator (D), enabling us to subject the entire man-machine system to sinusoidal oscillations in the sagittal plane. Due to the relatively low position of the platform rotation axis, its

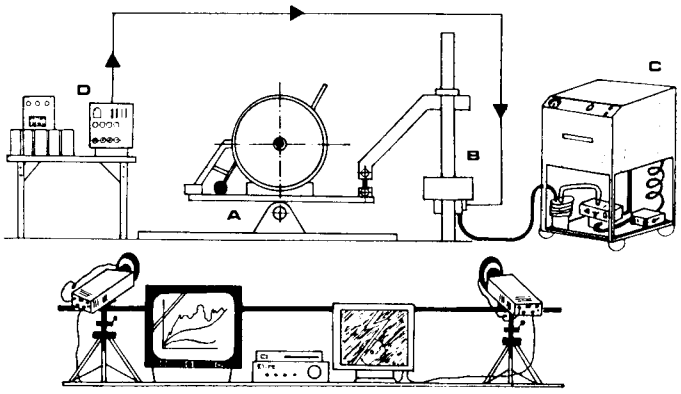

Figure 1.
Equipment for kinematic analysis of balance. (A: oscillating platform; B: hydraulic jack; C: hydraulic power supply; D: function generator).

distance from the various spinal measurement sites, and, as described later, the reduced angle rotation range, the main acceleration parameter taken into account was the horizontal (antero-posterior) component. This corresponds to dynamic conditions close to those obtained with a true antero-posterior translation movement generator.

The computerized video-based movement analyzer was an ELITE system, equipped with two video cameras. The movements of passive markers fixed on the subjects were recorded, while processing the tridimentional kinematic data. The system represents the functional pattern in harmonious groups of points bound by segments. The tracking procedure and kinematic analysis over a time-course provided us with data on linear and/or angular displacements, velocity and acceleration, and reconstruction of the trajectories of an unlimited number of markers. The sampling frequency was 100 Hz.

Procedure

The subject sat in a wheelchair fixed to the oscillating platform. He was fitted with four reflecting markers placed along the articular truncal axis as shown in **Figure 2**: at the temporal bone (M1), the superior head rotation center (M2), the inferior head rotation center (M3), and the iliac crest (M4). The two markers at the head rotation centers were placed according to the model of Berthoz (1983). Two other markers completed this model, the first on the rotation axis of the platform (M6) and the second (M5) at a vertical distance (h_0) of 20 cm from this axis.

The extent of acceleration transmitted to the subject depends on three main parameters: angular amplitude (a),

Figure 2.
Subject equipped with markers in experimental situation. (M1: temporal bone; M2: maxillar-superior head rotation center; M3: inferior head rotation center; M4: iliac crest; M5: oscillating platform; M6: platform rotation axis).

frequency (f) of the platform oscillations, and distance (h) to the platform oscillation axis.

Maximum tangential acceleration value is given by:

$$A_{max} = 4\pi^2 \cdot ahf^2$$

Parameter values were carefully set to meet with the following conditions: 1) to produce, at the head level, a main acceleration component in the antero-posterior direction, which requires relatively low platform rotation angle amplitude and 2) to subject the head to various acceleration levels corresponding to those reached in usual daily life situations, up to dynamic levels encountered in sport activities, while still remaining well below risky levels.

Four oscillation levels A,B,C, and D were thus defined:

A. $a = \pm 3.5°$ and $f = 1$ Hz which, for $h = 1$ meter (reference value), corresponds to a maximum acceleration of 2.4 ms^{-2}
B. $a = \pm 3.5°$ and $f = 1.25$ Hz, maximum acceleration 3.8 ms^{-2}
C. $a = \pm 5.5°$ and $f = 1$ Hz, maximum acceleration 3.8 ms^{-2}
D. $a = +5.5°$ and $f = 1.25$ Hz, maximum acceleration 6 ms^{-2}.

Despite different amplitude and frequency values, oscillation levels B and C led to the same acceleration, which allowed us to estimate the relative effects of these parameters on balance control.

For each stress level and at each marker, the maximal amplitude of acceleration was measured with a computerized optoelectronic movement analyzer. The following postural conditions were required of the subjects: hands on handrim, feet on the footrest, back leaning on the backrest, and visual axis oriented horizontally forward. Each test lasted 15 seconds, and the kinematic data were only recorded during the last 5 seconds.

Subjects

Two groups of six paraplegic subjects were selected (**Table 1**): one group was composed of high paraplegic athletes (HPA), with neurological levels between T4 and T8, and one group was composed of low paraplegic athletes (LPA), with neurological levels between T11 and L5.

The subjects were all members of the *Fédération Française Handisport* and the *Montpellier Club Handisport*, and did not have any spinal immobilization by mechanical means or bony fusion.

A group of six normal healthy athletes (NHA) was also used to provide us with a functional reference, and to assess variability in the normal balance process.

Damping Factor Definition

The maximum acceleration value as computed above corresponds to a theoretical situation where the marker, placed at a distance (h) on the rotation axis of the platform, is supposed to be attached to this axis by an infinitely rigid link.

Physiological reality is of course (and fortunately) quite different from this rigid model. Active and passive

Table 1:
Anthropometric and functional characteristics of the subjects.

Subj	age yrs	wgt kg	hgt cm	SCI lev	lesion	type	inj date	sport
HPA								
RF	30	55	170	T5	compl	spastic	1983	basket
AT	29	60	170	T4–T6	compl	spastic	1983	tennis
BA	31	71	185	T4	compl	spastic	1978	tennis
RP	33	66	177	T8	compl	spastic	1980	tennis
CL	30	55	168	T5	compl	spastic	1983	tennis
RA	37	70	178	T5	compl	spastic	1986	tennis
M	31.6	62.8	174.6					
SD	2.94	7.19	6.50					
LPA								
BP	27	70	180	L3	incompl	flacid	1984	tennis
BL	30	53	163	T12	incompl	spastic	1978	tennis
GS	24	63	170	T12	compl	spastic	1988	tennis
BS	29	78	184	T11	compl	spastic	1980	racing
GM	28	70	182	T12	compl	flacid	1988	tennis
NE	32	77	178	T12–L1	compl	flacid	1988	basket
M	28.3	68.5	176.1					
SD	2.73	9.35	8.06					
NHA								
VG	17	60	175	—	—	—	—	tennis
NG	21	80	191	—	—	—	—	tennis
BC	21	68	176	—	—	—	—	tennis
JD	27	60	165	—	—	—	—	tennis
CD	24	70	179	—	—	—	—	tennis
CR	28	65	173	—	—	—	—	tennis
M	23	67.1	176.5					
SD	4.14	7.49	8.52					

HPA = high paraplegic athletes; LPA = low paraplegic athletes; NHA = normal healthy athletes; SCI lev = spinal cord injury level; M = mean; SD = standard deviation; compl = complete; inj = injury

damping effects of the musculoskeletal components contribute to minimize acceleration at the head.

A damping factor (d) was determined to assess the subject's ability to maintain the head at the lowest level of acceleration. This factor is defined as the ratio of the real measured acceleration A_{meas} at a given site, to the theoretical acceleration A_{th} calculated at this site, assuming that the system is infinitely rigid:

$$d = A_{meas}/A_{th}$$

At a given point, the theoretical acceleration is proportional to the distance h to rotation axis. Consequently, if A_0 is the measured acceleration at a fixed reference point, situated at a distance h_0 from the axis of the oscillating platform, the theoretical acceleration A_{th} at any given point of the body, situated at distance h from the axis, can be calculated as:

$$A_{th} = A_0 \cdot h/h_0$$

Theoretically the value of the damping factor is between 0 and 1. The closer its value is to 0, the higher the damping effect at the considered site.

Moreover, the individual balance strategies can be characterized by assessing the relative contribution to damping of the cervical and thoracic spine. This can be achieved by calculating the difference between the values of the damping factor at the extremities of each spinal segment.

If A_3 is acceleration at the inferior head rotation center and A_4 acceleration measured at the iliac crest, the corresponding damping factors are:

$$d_3 = h_0/A_0 \cdot A_3/h_3 \text{ and } d_4 = h_0/A_0 \cdot A_4/h_4$$

The contribution to damping C_t of the thoracic spine is:

$$C_t = d_3 - d_4 = h_0/A_0(A_3/h_3 - A_4/h_4)$$

C_t, which is proportional to the difference of normalized accelerations A_3/h_3 and A_4/h_4 measured at each extremity of the thoracic segment, is fully representative of the damping effect of the considered spinal part. In the same way, we evaluated the contribution C_c to damping of the cervical spine $C_c = d_2 - d_3$. The damping factor was calculated for each oscillation level A, B, C, and D.

Data Analysis

A mean comparison test was selected to investigate the relationship between damping factor values and the extent of medular lesion and to define the relative contributions of the cervical and thoracic spine for maintenance of balance. The data were presented as the means ± standard deviation, with a significance level of $p < 0.05$.

RESULTS

Damping Factor at the Head

Statistical analysis of the values of the damping factor at the head did not lead to significant results. Nevertheless, graphical representation of values of the damping factor at the head (M1 marker) for oscillation levels A, B, C, and D and for HPA, LPA, and NHA groups (**Figure 3**), showed (except for very low oscillation levels), a marked decrease according to the neurological level. Moreover, this factor appears globally decreasing with the intensity of oscillation level. For the LPA and NHA subjects, the decrease in the damping factor was slightly perceptible in oscillation levels A, B, and C; whereas, it was more marked and regular in the HPA group under all conditions.

Cervical and Thoracic Spine Involvement in Balance

As shown in **Figure 4**, the damping factor decreased from iliac crest to the head. Nevertheless, the decreasing mode markedly differed according to the group considered. LPA and NHA showed a steadily progressive decrease at all oscillation levels, while there was a sharp decrease in the damping factor in the HPA group between M4 and M3, even at the lowest level.

From the damping factor values along the articular axis of the spine, we determined the contribution of the

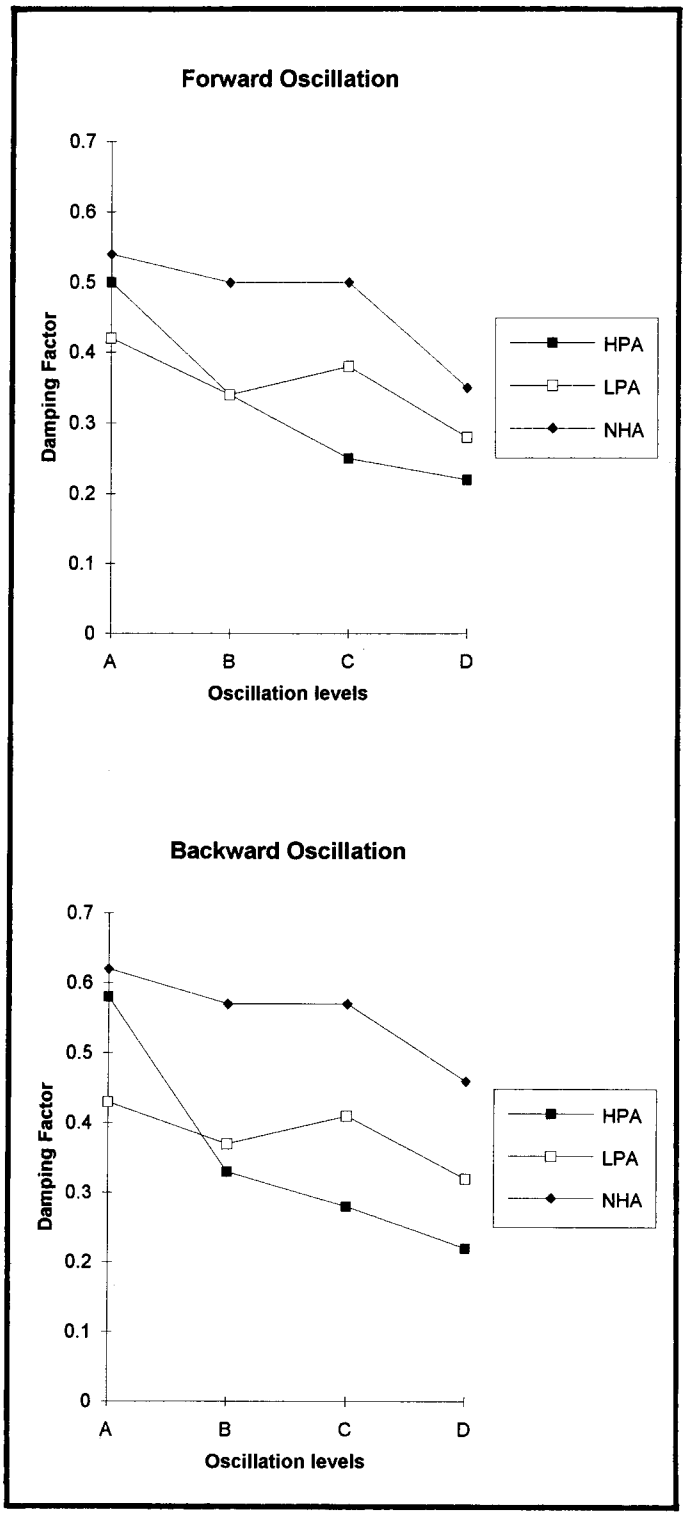

Figure 3.
Damping factor at the head (M1) during forward and backward oscillations for oscillation levels A, B, C, and D for high paraplegic athletes (HPA), low paraplegic athletes (LPA), and normal healthy athletes (NHA).

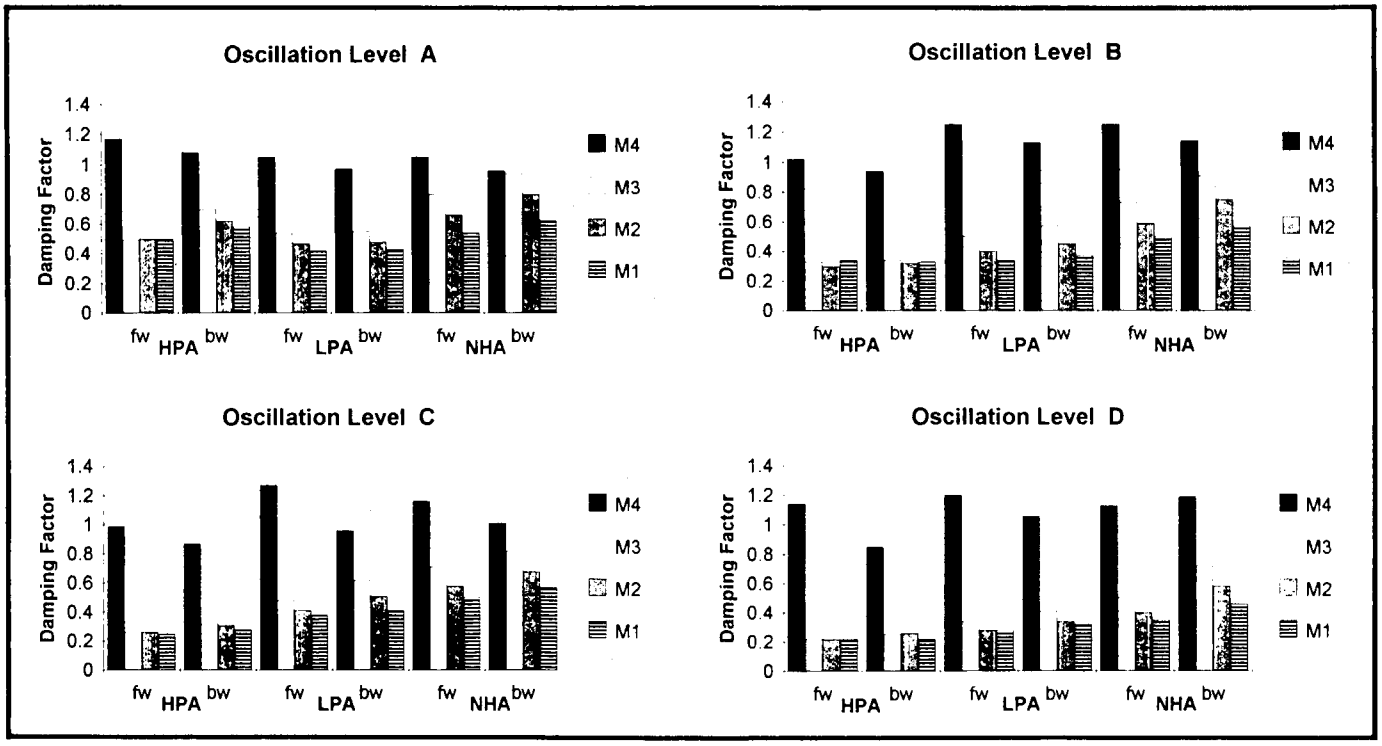

Figure 4.
Damping factor in forward (fw) and backward (bw) oscillations for high paraplegic athletes (HPA), low paraplegic athletes (LPA), and normal healthy athletes (NHA) groups for oscillation levels A, B, C, and D. A: 3.5°–1 Hz. B: 3.5°–1 Hz. 25 C: 5.5°–1 Hz. D: 5.5°–1.25 Hz. M4: Iliac crest. M3: Inferior rotation center. M2: Superior rotation center. M1: Temporal.

cervical and thoracic spine to balance. Regarding the contribution of the thoracic spine during forward oscillations (**Table 2a**), we observed significant differences between HPA and NHA for oscillation level A ($p < 0.05$). There were no marked differences between the three groups during the backward oscillation phase.

Regarding the contribution of the cervical spine to balance during the forward oscillation phase (**Table 2b**), we observed significant differences between HPA and LPA for level B, between LPA and NHA for level C, and more generally, between HPA and NHA for levels A, B, C, and D ($0.01 < p < 0.05$). Regarding the contribution of the cervical spine to balance during backward oscillations, we observed significant differences between HPA and LPA for level C, between LPA and NHA for level A, and between HPA and NHA for levels B, C, and D.

DISCUSSION

Variations in the Damping Factor at the Head

On the whole, we observed that the damping factor values at the head decreased with the intensity of stress. This decrease was barely perceptible for NHA and LPA under moderate stress (oscillation levels A, B, and C). Subjects of the two groups with sufficient or normal physical capabilities did not have any particular difficulties in controlling balance. During oscillation levels B and C, nearly identical damping factor values were obtained at the head. This is normal considering, as before, that accelerations applied in these two tests were identical, despite the different amplitude and frequency values.

The behavior of HPA differed markedly, with an increase in the damping effect for the higher mechanical excitation amplitudes. Moreover, in this group we noted a regular decrease in the damping factor with increasing intensity of oscillation. The relatively high damping factor value (0.50) during oscillation level A could be explained by the subject's search for the best strategy to obtain the most efficient stabilization of the head. For more intense stress (oscillation levels B, C, and D), there was a relation between the damping factor value and the neurological level of the subject.

We also observed that in all conditions, normal subjects had a damping factor value higher than that of the two groups of paraplegic subjects. This could reflect a reduced

Table 2a:
Contribution to damping of the thoracic spine by oscillation level.

Oscillation			Level			
			A	B	C	D
Forward						
Contribution	HPA	M	0.68	0.69	0.73	0.92
		SD	0.23	0.19	0.10	0.28
	LPA	M	0.51	0.75	0.79	0.86
		SD	0.19	0.24	0.26	0.13
	NHA	M	0.25	0.52	0.47	0.62
		SD	0.25	0.28	0.28	0.24
Comparison	HPA/LPA	t	1.25	0.46	0.45	0.43
		p	—	—	—	—
	LPA/NHA	t	1.85	1.39	1.87	1.96
		p	—	—	—	—
	HPA/NHA	t	2.82	1.09	1.97	1.81
		p	< 0.05	—	—	—
Backward						
Contribution	HPA	M	0.39	0.60	0.55	0.60
		SD	0.34	0.36	0.18	0.25
	LPA	M	0.42	0.56	0.35	0.65
		SD	0.22	0.37	0.23	0.29
	NHA	M	0.02	0.23	0.18	0.47
		SD	0.41	0.34	0.34	0.42
Comparison	HPA/LPA	t	0.2	0.15	1.54	0.29
		p	—	—	—	—
	LPA/NHA	t	1.91	1.49	0.89	0.79
		p	—	—	—	—
	HPA/NHA	t	1.52	1.66	2.13	0.59
		p	—	—	—	—

HPA = high paraplegic athletes; LPA = low paraplegic athletes; NHA = normal healthy athletes; M = mean; SD = standard deviation; t = t value; p = significance

need for active damping of normal subjects who had integral sensori-motor potential. This was confirmed by particularly high damping factor values (0.6) in the low stress oscillation levels A, B, and C. It can be assumed that for HPA and LPA, the higher the neurological level, the more sensitive the subjects were to mechanical perturbations. Consequently, they tried to minimize the effect of the mechanical perturbation by minimizing the amplitude of the head acceleration. On the contrary, the nondisabled athletes showed less intense reactions to mechanical perturbations and were less sensitive to the conditions. All of the above observations were valid for the forward and backward stresses. Nevertheless, the damping factor value was noticeably higher during backward stresses. This phenomenon may be related to the decrease in the number of degrees of freedom of the thoracic and lumbar articular spine during the contact phase of the back with the backrest of the wheelchair. In the next section, we analyze the involved articular and muscular structures in order to determine a satisfactory damping effect at the head.

Balance Strategy: Thoracic and Cervical Spine Involvement

During forward stress, the comparison between HPA and NHA behavior in all tests showed significant differences of the contribution of the cervical spine to balance.

Table 2b:
Contribution to damping of the cervical spine by oscillation level.

Oscillation			Level			
			A	B	C	D
Forward						
Contribution	HPA	M	0.00	0.00	0.00	0.00
		SD	0.13	0.09	0.09	0.08
	LPA	M	0.11	0.16	0.10	0.05
		SD	0.01	0.11	0.07	0.10
	NHA	M	0.26	0.22	0.21	0.15
		SD	0.10	0.13	0.04	0.06
Comparison	HPA/LPA	t	1.68	2.62	1.81	1.10
		p	—	< 0.05	—	—
	LPA/NHA	t	2.24	0.86	3.08	1.85
		p	—	—	< 0.02	—
	HPA/NHA	t	3.68	3.32	4.86	3.50
		p	< 0.01	< 0.02	< 0.01	< 0.01
Backward						
Contribution	HPA	M	0.12	0.004	0.03	0.03
		SD	0.20	0.18	0.18	0.10
	LPA	M	0.12	0.20	0.20	0.09
		SD	0.11	0.15	0.08	0.12
	NHA	M	0.32	0.34	0.25	0.26
		SD	0.14	0.15	0.12	0.16
Comparison	HPA/LPA	t	0.02	0.81	2.61	0.80
		p	—	—	< 0.05	—
	LPA/NHA	t	2.49	1.51	0.78	1.87
		p	< 0.05	—	—	—
	HPA/NHA	t	1.81	3.20	2.96	2.68
		p	—	< 0.02	< 0.02	< 0.05

HPA = high paraplegic athletes; LPA = low paraplegic athletes; NHA = normal healthy athletes; M = mean; SD = standard deviation; t = t value; p = significance

Nevertheless, a noticeable difference ($p < 0.05$) was obtained for the B oscillation level (favoring frequency of oscillations) between HPA and LPA groups.

Concerning the contribution of the thoracic spine, the only significant difference appeared between HPA and NHA during oscillation level A. During backward stress, we observed the same type of results as for forward stress. There were significant differences between HPA and NHA during oscillation levels B, C, and D. However, there was a marked difference ($p < 0.05$) between HPA and LPA for the oscillation level C (favoring amplitude of oscillation). No significant difference was observed between the three groups of subjects for thoracic spine involvement in balance.

Based on the obtained data, the following characteristic behaviors for subjects of the three groups can be presented:

- The involvement of the thoracic spine in HPA and LPA increased with the intensity of stress and was markedly higher than that of NHA for both forward and backward stresses
- The involvement of the cervical spine in balance in HPA was zero or very weak during both forward and backward stresses
- All subjects showed a tendency to fix the cervical spine when the stress increased.

The neck stiffening tendency and consequent decrease in the mobility of the cervical spine clearly appeared in HPA and, to a lesser degree, in LPA. This stiffness was probably obtained by intense muscular action, generating fatigue and considerable mechanical articular constraints at this level. Similarly, spinal immobilization from orthotic devices or vertebral fusion, also causing spine rigidification, might have direct consequences on the transmission of acceleration. It would be interesting to test the method on a population of subjects with spinal immobilization at various levels.

In addition, it appears that the neurological level was not completely related to the damping factor values and to the balance strategies of the paraplegic subjects. Indeed, we observed less stiffness for two subjects of the thoraco-neurological level group than for one subject of the lumbar neurological level group. In contrast to other wheelchair athlete assessments, while we observed a close correlation between neurological level and biological responses (8,10), the present study demonstrates that balance strategies are more difficult to characterize.

We consider that the orientation and training of paraplegic and tetraplegic wheelchair athletes will require association of different functional evaluation parameters. Adaptation of physical activities to the capacities of the subjects demands more than knowledge of the neurological level. A multiparametric approach to athletic behavior, with this kind of balance evaluation, offers a means to optimize capacities while preventing pain and traumatism related to sport activities.

CONCLUSION

In spite of the small subject sample size, this study reveals the balance capabilities of paraplegic athletes. It also offers a means of analyzing their behavior under well-defined mechanical conditions. There seems to be a relationship between the damping factor measured at the head and the subject's neurological level, which indicates the importance of this parameter. This factor could be a relevant quantitative indicator for assessing the ability of the paraplegic subject to obtain efficient body balance in the sitting position. For instance, this factor might be taken into consideration before orienting subjects toward particular sport activities requiring good control of sitting postures under highly dynamic stress. The damping factor could also be used in wheelchair design to evaluate the effect of given fittings on balance.

Balance strategies can be described through analysis of the relative contribution to balance of the cervical and thoracic spine. There seemed to be a relationship between the degree of stiffening of the cervical spine and the subject's neurological level. Cervical spine stiffening, generally obtained by violent contractions of the neck muscles, could be the cause of spinal impairment. It thus seems reasonable to assume that it is better to be among those with less neck stiffening.

The proposed method seems effective in quantifying individual postural balance and estimating the risk of practicing particular sport activities. Nevertheless, considering the small number of athletes tested, the complexity, and thus the variability of the balance process, some differences in subject behavior were not clearly seen. Furthermore, the observed lack of significant differences according to neurological levels indicates that knowledge of this clinical factor alone cannot determine real individual ability to maintain balance, and points out the need for complementary quantitative assessment techniques. More precise characterization of the balance strategies will require further research on a wider sample of subjects. This could be achieved by analyzing the synergic activity of cervical muscles during the voluntary activity of wheelchair propulsion. We will attempt to establish relationships between neck reflex stiffening and development of pain or musculo-skeletal traumatisms in the cervical spine area.

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REFERENCES

1. Brubaker CE, McClay I. Determination of the relationship of mechanical advantage on handrim propulsion efficiency. In: Wheelchair mobility. Charlottesville, VA: University of Virginia Rehabilitation Engineering Center, 1983:1–7.
2. Brubaker CE, McClay I, McLaurin CA. The effect of mechanical advantage on handrim propulsion efficiency. In: Proceedings of the 2nd International Conference on Rehabilitation Engineering, June 17–22, Ottawa, ON, 1984.
3. Glaser RM, Sawka M, Wilde S, Woodrow B. Energy cost and cardiopulmonary responses for wheelchair locomotion and walking on tile and on carpet. Paraplegia 1981:19:220–6.
4. Van Der Woude LHV, Veeger DJ, Rozendal RH. Ergonomics of manual wheelchair propulsion. In: Wheelchairs research, evaluation and information. Milano, Italy: COMAC BME, 1988:101–12.

5. Pendergast D. Cardiovascular, respiratory and metabolic responses to upper body exercise. Med Sci Sports Exerc 1989:21(5):121–5.
6. Sawka MN. Physiology of upper body exercise. Exerc Sports Sci Rev 1986:14:175–211.
7. Sawka MN, Latzka WA, Petolf KB. Temperature regulation during upper body exercise: able bodied and spinal cord injuries. Med Sci Sports Exerc 1989:21(5):97–100.
8. Coutts KD, Rhodes EC, McKenzie DC. Sub-maximal exercise responses of tetraplegics and paraplegics. J Appl Physiol 1983:55(2):479–82.
9. Glaser RM. Arm exercise training for wheelchair users. Med Sci Sports Exerc 1989:21(5):149–57.
10. Van Loan MD, McClur S, Luftin JM. Comparison of physiological responses to maximal arm exercise among able-bodied, paraplegics and quadriplegics. Paraplegia 1987:25(5):397–405.
11. Rozendal RH, Van Der Woude LHV, Veeger HRJ. L'ergonomie et la conception des moyens mobiles. In: Ergonomie et Handicaps Moteurs. Paris: CTNERHI, 1988:97–113.
12. Sanderson DJ, Sommer HJ. Kinematic features of wheelchair propulsion. J Biomech 1985:18(6):423–9.
13. Tupling SJ, Davis GM, Pierrynowsky MR, Shepard RJ. Arm strength and impulse generation: initiation of wheelchair movement by the physically disabled. Ergonomics 1986:29(2):303–11.
14. Brubaker CE., McLaurin CA, Gibson JD, Stoots TH. Seat position and wheelchair performance. In: Wheelchair mobility. Charlottesville, VA: University of Virginia Rehabilitation Engineering Center, 1981:76–81.
15. Van Der Woude LHV, Veeger DJ, Rozendal RH, Sargeant TJ. Seat height in handrim wheelchair propulsion. J Rehabil Res Dev 1989:26(4):31–50.
16. Veeger DJ, Van Der Woude LHV, Rozendal RH. The effect of rear wheel camber in manual wheelchair propulsion. J Rehabil Res Dev 1989:26(2):37–46.
17. Burnham RS, May L, Nelson E, Steadward R, Reid DC. Shoulder pain in wheelchair athletes: the role of muscle imbalance. Am J Sports Med 1993:21(2):238–42.
18. Umbach I, Heilporn A. Usure prematuree au niveau du rachis cervical et des membres superieurs chez le paraplegique. Ann Reeducation Med Phys 1989:32:769–71.
19. Bayley JV, Cochran TP, Sledge CB. The weight bearing shoulder: the impingement syndrome in paraplegics. J Bone Joint Surg 1987:69(5):676–8.
20. Ferrara M, Davis R. Injuries to elite wheelchair athletes. Paraplegia 1990:28:335–41.
21. Hauptmann HW, Littman BH. Rheumatologic considerations in spinal cord injury. Phys Med Rehabil 1987:1:389–99.
22. Nadeau G, Pailler D, Wirotius JM, Tronca R, Benichou M. L'epaule du sportif en fauteuil roulant. Kinesiology 1986:25:195–8.
23. Cooper RA. An exploratory study of racing wheelchair propulsion dynamics. Adapt Phys Act Q 1990:7(1):74–85.
24. Coutts KD. Kinematics of sport wheelchair propulsion. J Rehabil Res Dev 1990:27(1):219–6.
25. Dauphin M. Deambulation du paraplegique en fauteuil: mouvements de la tete et du tronc (Thesis). Diplome d'Etude Approfondie en Sciences et Techniques Aplliquees au Handicap et a la Readapation, Universite de Bourgogne, Faculte de Medecine de Dijon, 1992.
26. Bernard PL, Chollet D, Peruchon E, Micallef JP. Contraintes cervico-rachidiennes et paraplegie. Sci Sport 1990:5:113–27.
27. Bernard PL. Recherche d'une methode d'evaluation des capacites musculaires, energetiques et posturales de sportifs paraplegiques en fauteuil roulant (Thesis). Montpellier, France: University of Montpellier, 1992.

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A survey of marginal wheelchair users

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Abstract—Significant numbers of wheelchair users experience difficulties with propulsion due to impaired upper limb function (termed marginal users for this study). A survey of wheelchair users in Tayside, Scotland, was carried out to identify and describe the marginal user population and their propulsion difficulties. Subjects for the survey were identified from the records of National Health Service wheelchair users at Dundee Limb Fitting Centre. Subjects were interviewed at home about their wheelchair-propelling experiences.

Survey results indicated that marginal users represent approximately 15% of the occupant-propelled wheelchair population. The average age of the marginal users surveyed was 48 years and the modal diagnosis was multiple sclerosis. Fifty-nine percent of the marginal users questioned felt that their wheelchairs were not adequate for their requirements.

Key words: *marginal users, multiple sclerosis, propulsion, wheelchair.*

INTRODUCTION

Wheelchair wheel position and other variables, such as castor type and size, handrim type, wheel camber, and backrest angle, can affect the efficiency and effectiveness of wheelchair propulsion for strong, fit users (1–6). However, the effects of such variables on the propulsion of less able users, who are only just capable of self-propulsion, have been neglected. Such wheelchair users are termed marginal users. The effect of optimizing the wheel position on an athlete's wheelchair may be a slightly faster time in a race, but for a less able, marginal wheelchair user it may be the difference between dependent or independent propulsion. The marginal user, therefore, stands to gain more from correct wheelchair adjustment.

The Scottish Office Home and Health Department funded an 18-month research project entitled "The Determination of Optimum Wheel Configurations for Wheelchair Users." The aim of the project was to determine the influence of wheel configuration on manual wheelchair propulsion. These influences were highlighted in clinical trials of marginal wheelchair users. Before the trials could take place, it was necessary to identify and describe the marginal wheelchair user population. A survey of wheelchair users in Tayside was completed for this purpose. This paper describes the survey and its results.

METHOD

Subject Selection

The survey of marginal users began with the identification of potential subjects from the records of over 3,000 wheelchair users (the total wheelchair population of Tayside) at Dundee Limb Fitting Centre. Three selection criteria were applied at this stage:

1. Self-propelling wheelchair users only were considered.
2. A geographical limit was imposed to limit travel. Users were selected only if they lived in the cities of Dundee and Perth or on the coastal strip of Angus (66 percent of the Tayside population).
3. Pre-school users were not considered; at the time of the survey, no children under 5 years old living in Tayside were self-propelling.

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A list of 700 names drawn from the first stage was passed on to the community occupational therapy service and to the staff at specialist educational, residential, and vocational centers for a further phase of sifting. The following exclusion criteria were applied at this stage:

1. Foot/feet only and double rim (one arm) propellers were excluded, as the project dealt specifically with two-wheel propulsion.
2. Users with good upper limb function were excluded. In some cases, impaired upper limb function was the result of problems other than the primary diagnosis (e.g., lower limb amputees with cardiovascular problems).
3. Very frail elderly people were not included due to the demanding nature of testing.
4. Users with poor motivation were excluded due to the lengthy and demanding nature of the trials.
5. Users with poor communication skills were excluded, as the trials made extensive use of subjective user feedback.

One hundred and seventeen users remained from the original list and were contacted by letter inviting them to discuss their propelling experiences. Eighty-three agreed to this request and were interviewed by the project staff.

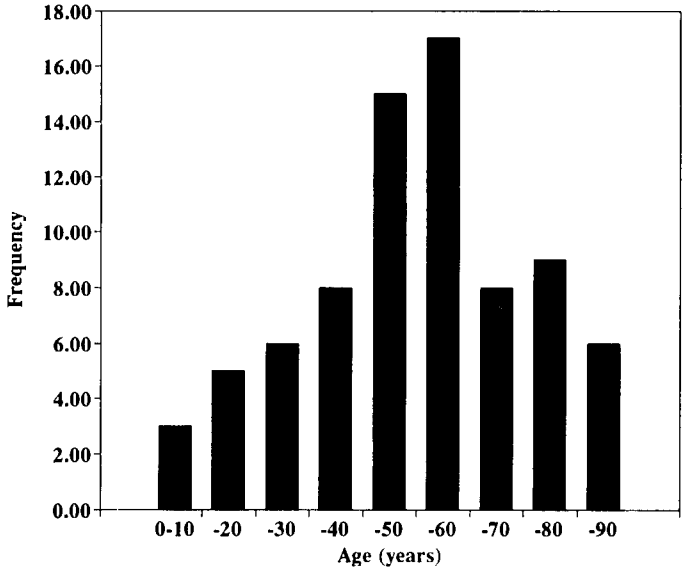

Figure 1.
Age distribution of marginal users.

Interview and Questionnaire

Subjects were interviewed in their most familiar wheelchair environment. This was usually at home, although some children were seen in their school. This gave the interviewer the opportunity to see wheelchair propulsion under normal conditions. The interview, which took the form of a questionnaire, was designed to highlight factors limiting successful wheelchair propulsion and daily use. Particular reference was made to wheel configuration. Propulsion difficulties were divided into categories of technical, functional, and environmental. Information was requested on the following topics:

- User background detail, including information on support services
- User medical background and diagnosis
- Wheelchair and seating information, including propulsion technique and ability
- Wheelchair environments
- Daily wheelchair activities, including transfer technique.

Of those interviewed, 3 were not considered to be sufficiently impaired. This left a group of 80 marginal users (44 male, 36 female).

RESULTS AND DISCUSSION

Estimation of the Marginal User Population Size

The number of marginal users identified in the survey was scaled to account for the subjects excluded in the sampling process (geographical exclusions, frail elderly, and users with poor motivation and communication) and to provide a minimum estimate for the population size of marginal wheelchair users. This gave an estimated total of 145 marginal users in Tayside, representing approximately 15 percent of the occupant-propelled wheelchair population.

The populations of Tayside, Scotland and the UK are approximately 400,000, 5 million, and 55 million respectively. Tayside has a sufficiently large population to be considered representative, in general, of the UK population. Approximate estimates, therefore, of Scotland and the UK's marginal user population sizes are 1,800 and 20,000.

Description of Marginal User Population

Twenty-four percent (19) of the marginal wheelchair users studied came from the 0–30-year age group (**Figure 1**). Typical diagnoses for these young users were spina bifida, cerebral palsy, and muscular dystrophy.

Fifty percent (40) of the users studied were between 30 and 60 years of age (**Figure 1**). This was reflected by many of the diagnoses being associated with deterioration in middle age: multiple sclerosis 26 percent (21), rheumatoid arthritis, 9 percent (7), amputations, 6 percent (5), and cerebral vascular accident (CVA), 6 percent (5) (**Figure 2**).

Twenty-six percent (21) of the sample were in the 60–90 age group, many of whom had been reasonably active wheelchair users in the past, but had become marginal through the combined effects of ageing and their diagnoses.

Figure 2 shows users by category. The largest diagnostic categories of marginal users in Tayside are multiple sclerosis, 26 percent (21), spina bifida, 10 percent (8), rheumatoid arthritis, 9 percent (7), and cerebral palsy, 8 percent (6).

As shown in **Figure 3**, 75 percent (60) of the subject group were using standard Ministry model 8 wheelchairs,

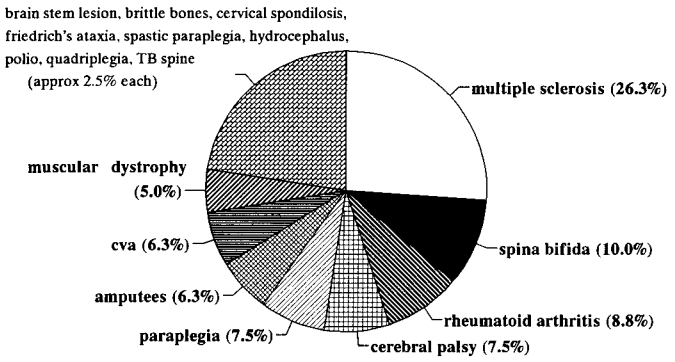

Figure 2.
Diagnoses of marginal users.

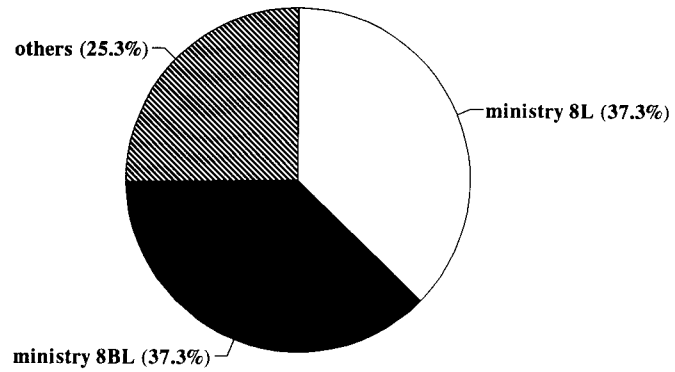

Figure 3.
Wheelchair types employed by marginal users.

Figure 4a.
Wheelchair Model 8L.

such as models 8L, 8BL, 8LC, and 8LJ. **Figures 4a** and **4b** illustrate examples of the most-used models (8L and 8BL). The remaining 25 percent (20) were using alternative models: Carter's Activ (4), Carter's Imperial (1), Chevron (1), Everest & Jennings (1), Newton (4), Poirier Roller (1), Quickie (2), Quickie Breezy (1), Remploy Roller (1), Swede Champ (2), Swede Elite (1), Vessa Variant (1). Of these alternative models, 15 percent (3) were supplied

Table 1.
Comments and different types of wheelchairs.

Comment	Model 8 (60)		Others (20)		Both (80)	
Wheelchair inadequate	67%	(40)	35%	(7)	59%	(47)
Technical problems	33%	(20)	10%	(2)	28%	(22)
General discomfort	42%	(25)			31%	(25)
Castors too small	23%	(14)			18%	(14)
Poor drive wheel position	15%	(9)			11%	(9)
Heavy to propel	18%	(11)	10%	(2)	16%	(13)
High rolling resistance	15%	(9)	5%	(1)	13%	(10)
Obtrusive armrests	13%	(8)	5%	(1)	11%	(9)
Obtrusive footplates	7%	(4)	15%	(3)	9%	(7)
Poor backrest angle	8%	(5)	5%	(1)	8%	(6)
Handrim grip too narrow	8%	(5)			6%	(5)
Uncomfortable seating	5%	(3)	5%	(1)	5%	(4)

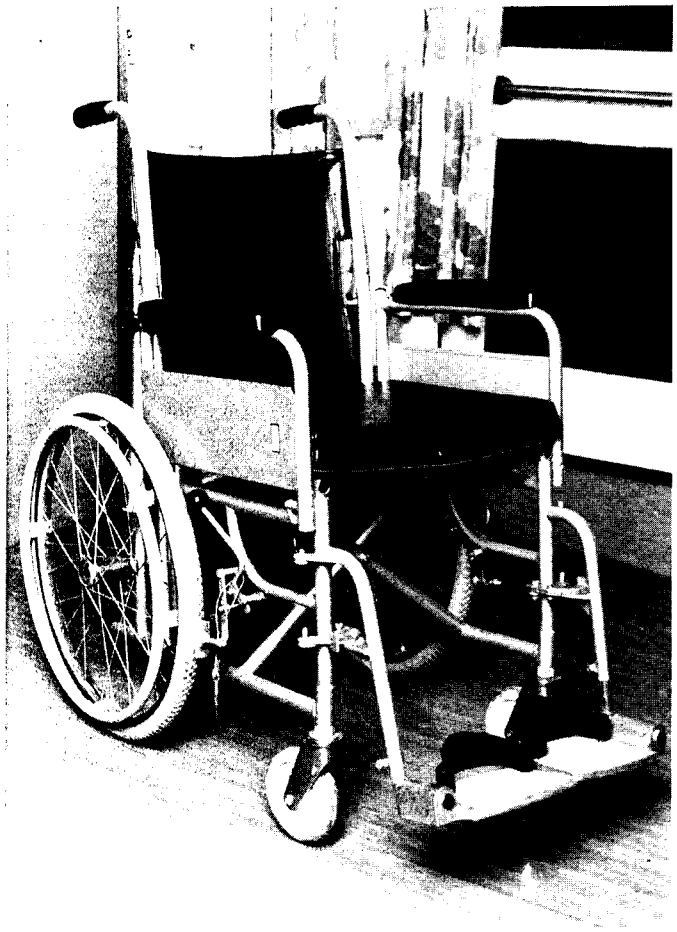

Figure 4b.
Wheelchair Model 8BL.

through the National Health Service (NHS) and 85 percent (17) were purchased privately as the users felt that the NHS prescriptions were inadequate. (**Figure 3** and **Table 1**).

Only 39 percent (31) of the marginal users questioned propelled with the rims only, 54 percent (43) gripped the tire and rim together and 7 percent (6) the tire only. This

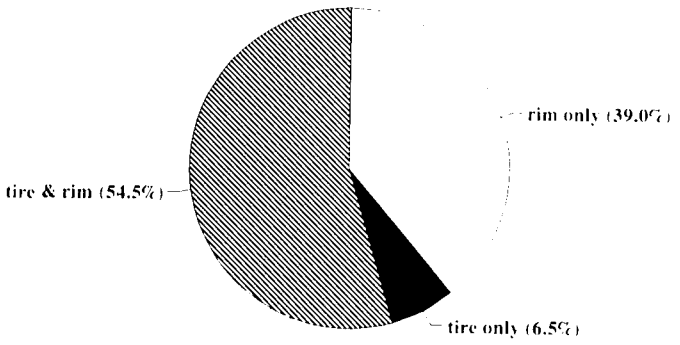

Figure 5.
Wheel grip methods of marginal users.

suggests that for 60 percent of those questioned, the handrims did not fulfill the purpose for which they were designed (**Figure 5**).

Fifty-five percent (44) of users transferred by standing and pivoting on their feet, reflecting the fact that marginal users often have some limited lower limb function. Only 23 percent (18) employed the sideways transfer technique, as most did not have sufficient upper limb strength for lifting themselves over the wheel (**Figure 6**).

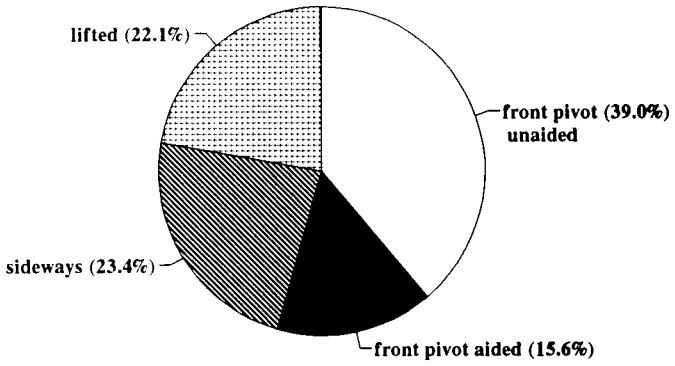

Figure 6.
Transfer techniques of marginal users.

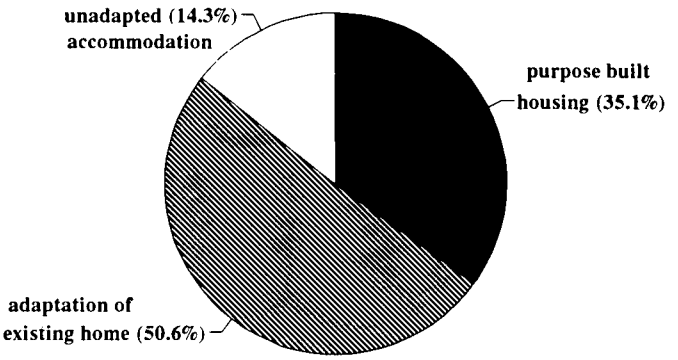

Figure 7.
Home environments of marginal users.

Nearly two thirds, 65 percent (52), of those interviewed did not live in purpose-built wheelchair housing (**Figure 7**). This lead to further propulsion difficulties caused by environmental factors. Standard width doorways caused access problems, particularly when turning through them in wheelchairs with footrests attached. Ramps created problems with rearward tipping instability. Door sills were reported to be difficult to negotiate and caused rearward tipping instability on impact with the front castor wheels. Inappropriate floor covering caused high rolling resistance.

Each user was asked to comment on the perceived adequacy of his/her wheelchair; 59 percent (47) felt that their chairs were inadequate with only 41 percent (33) feeling that their chairs were adequate. Those people who found their wheelchairs adequate often qualified this by saying that they had tried no other models.

Twenty-eight percent (22) of the wheelchair users interviewed thought that technical features of their wheelchairs inhibited successful propulsion and activities of daily living (e.g., poor wheel reach). Twenty-five other users (31 percent) were aware only of general discomfort and difficulties in using their chairs but were unable to describe the source of the problems.

Users were invited to give a subjective account of their own propelling experiences, identifying aspects of the wheelchair that they felt influenced propulsion performance. These comments were divided into the following categories:

Propulsion: Eighteen percent (14) commented that small casters compromised propulsion by creating rearward tipping instability when negotiating small obstacles such as door sills. Poor drive wheel position was noticed by 11 percent (9).

Rolling Resistance: Sixteen percent (13) felt that their wheelchairs were heavy to propel; 13 percent (10) thought that their castor type, together with floor surfaces, created high rolling resistance.

Wheelchair Frame: Eleven percent (9) commented that their armrests were obtrusive and inhibited wheel reach; 9 percent (7) said that their footplates were obtrusive and inhibited access through narrow spaces; and 8 percent (6) thought that their backrest angle compromised wheel reach and propulsion.

Handrims: Six percent (5) remarked that their handrims were too narrow to grip.

Seating: Five percent (4) found their wheelchairs to be uncomfortable.

Other Comments: Some users commented on more individual problems. These included:

- Large wheel diameters compromise sideways transfers
- Forward-positioned wheels compromise sideways transfers
- The plastic coating of the rims and the rubber of the tires caused allergic reactions
- Providing space for winter clothing widens the seat and inhibits wheel reach
- Cushion height inhibits wheel reach, particularly for someone with short arms
- Cushion height influences rearward tipping stability
- Punctures are difficult to cope with, particularly when living alone.

Table 1 compares comments about Ministry model 8 and other types of wheelchairs.

CONCLUSION

The survey results indicate that 15 percent (145) of the self-propelling wheelchair population in Tayside are experiencing difficulty in propelling their wheelchairs and may be termed marginal users. They exhibited the following characteristics:

1. Functional deterioration in middle age due to conditions such as multiple sclerosis, rheumatoid arthritis, amputation, and cerebrovascular accident (CVA); 50 percent are in the 30–60 age group.
2. Frailty due to old age creates propulsion problems for established wheelchair users; 26 percent are in the 60–90 age group.
3. Young marginal users (24 percent—under 30 years old) typically have conditions such as spina bifida, cerebral palsy, and muscular dystrophy.
4. Sixty-five percent of those questioned faced indoor and outdoor access problems as they lived in non-purpose-built accommodation.
5. The majority (59 percent) of users questioned said that their wheelchairs were inadequate for their requirements. Typical wheelchair problems included inadequate wheel positions (11 percent), castor wheels that were too small (18 percent), high rolling resistance (16 percent), obtrusive footplates (11 percent), and unsatisfactory handrims (6 percent).

The survey results provided valuable information about marginal wheelchair users. In addition, it was a valuable source of subjects for the subsequent investigation into propulsion by marginal users.

REFERENCES

1. Brubaker CE. Determination of the effects of seat position on propulsion performance. In: Wheelchair mobility 1983–1988. Charlottesville, VA: University of Virginia Rehabilitation Engineering Center, 1988:24–6.
2. Brubaker CE. A predictive model for wheelchair propulsion optimization. In: Wheelchair mobility 1983–1988. Charlottesville,

VA: University of Virginia Rehabilitation Engineering Center, 1988:1–9.
3. Van der Woude L. Manual wheelchair propulsion: an ergonomic perspective. Amsterdam: Free University Press, 1989.
4. Flynn CM. Determination of optimum seat/wheel position for users of self-propelled wheelchairs (Thesis). Dundee, Scotland, University of Dundee, School of Biomedical Engineering, 1989.
5. Rose A. An investigation of the influence of wheel position on the propulsion capabilities of wheelchair users (Thesis). Dundee, Scotland, University of Dundee, School of Biomedical Engineering, 1990.
6. Masse LC, Lamontagne M, Riain MD. Biomechanical analysis of wheelchair propulsion for various seating positions. J Rehabil Res Dev 1992:29(3):12–34.

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Distributed random electrical neuromuscular stimulation: Effects of the inter-stimulus interval statistics on the EMG spectrum and frequency parameters

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Abstract—An electrophysiological approach was used to study a distributed random electrical neuromuscular stimulation (ENMS) scheme in which a probability density is assigned to the inter-stimulus intervals (ISI) of the stimuli. One of the objectives of using ENMS techniques in the study of skeletal muscles is to obtain information about the electrical, physiological, and mechanical properties of muscles in a near-physiological situation under a well-controlled experimental design in which problems related to the uncertainty of firing patterns of the central nervous system and physiological interference are avoided. In particular, ISI with a Gaussian density were varied in mean rate, standard deviation (SD), and coefficient of variation. The influence of varying ISI, and the interaction of the ISI statistics with compound motor unit action potentials (CMUAP) on EMG power spectra and their frequency parameters, was assessed theoretically using a mathematical model which is similar to that of EMG signal generation in the electrophysiological case. In order to quantify the effects of ISI statistics on the EMG spectrum, the median frequency was calculated as a function of stimulation rate using analytical expressions for various values of the coefficients of a Gaussian ISI variation. The results obtained suggest that 1) the interaction between ISI statistics and the shape of the CMUAP plays a major role in determining the EMG spectrum; 2) the median frequencies (MF) determined from EMG spectra tend to increase with increasing mean rates of stimulation for a given CMUAP. The rate of increase of the MF depends on the coefficient of the ISI variation; 3) the EMG spectra of random electrically stimulated muscle show peaks at the mean rate of stimulation, and multiples of it, when the coefficient of variation of ISI is small. These peaks decrease in magnitude with increasing coefficients of variation of ISI; and, 4) a variation in the ISI should be introduced in the ENMS, when a reproduction of 'normal' EMG spectra is needed. These results are consistent with those reported for voluntary contraction of skeletal muscles.

Key words: *electrical neuromuscular stimulation, electromyography (EMG), spectral analysis.*

INTRODUCTION

Electrical neuromuscular stimulation (ENMS) has become a common and important technique to study muscle activities, including electromyographic signals (EMG), biomechanical outputs, such as muscular force and vibromyographic (VMG) events (known also as muscle sounds), and their relationships (1–7). One of the objectives of using ENMS techniques in the study of skeletal muscles is to obtain information about the electrical, physiological, and mechanical properties of muscles in a near-physiological situation under a well-controlled experimental design in which problems related to the uncertainty of firing patterns of the central nervous system and physiological interference are avoided. For studies with this objective, it is necessary to select appropriate stimulation parameters and strategies, in accordance with findings from research in electrophysiology.

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A literature review (8–13) of electrophysiological studies of muscle identifies the following features: 1) force production in skeletal muscle is controlled by the number of motor units and the rate of firing of each unit; 2) EMG signals from active units are uncorrelated during low and medium voluntary effort, but may become synchronized at high levels of effort; and, 3) firing of alpha motor neurons is a random process, in the sense that the time interval between two successive spikes, that is, the interspike interval (ISI), is a random variable described by a probability density function (pdf). These random firing patterns of motor units have been widely studied during voluntary contraction in human skeletal muscles (9,14–17). Effects of the firing statistics, especially the mean firing rate, on the EMG spectrum have been described using mathematical models and experiments during tasks of moderate, nonfatiguing, constant effort isometric contractions (15,18–22). These findings suggest that the choice of the ISI statistics, in a study involving ENMS, may influence the median and mean frequencies of the EMG spectrum, two parameters commonly used to assess muscle properties during tests which involve voluntary contractions (10,14,23).

Because of technical difficulties and a lack of adequate physiological information as well as because of the research questions being asked, early research involving ENMS often was not aimed at representing the actual physiological situation. In the past 20 years, researchers have become more and more interested in the electrical, mechanical, and physiological properties of muscle contraction, and they have started to incorporate electrophysiological findings into ENMS approaches. Working on the lower leg of the cat, Petrofsky described a computer-controlled stimulator and a special electrode array that could control the recruitment pattern of motor units during electrical stimulation (2). More recently, a sophisticated neuromuscular stimulation system was described that can be used to change forces in skeletal muscle by varying firing rates and recruitment control strategies (24). Using this system, Solomonow et al. systematically studied the power spectrum characteristics of the M-wave and the relation between force and EMG (5), and Baratta et al. (25) examined carefully the dependence of frequency response of muscle on control strategy.

To our knowledge, however, most early experimental designs using ENMS did not take into account the random nature of physiological motor neuron firing. Therefore, a potential gap exists between the research using periodic ENMS, and the findings obtained in electrophysiological studies of muscle. Physiological activation of motor units in intact skeletal muscle occurs through distributed random excitation. Distributed stimulation, in this context, means activating muscle according to the size principle, using multichannel stimulation. Random excitation refers to the random time interval between successive stimuli which may be described by a normal ISI distribution. Conventional electrical neuromuscular stimulation is typically performed using periodic or near periodic stimulation patterns. The coefficient of variation of the ISI of physiological and periodic activation of muscle is different: a coefficient of about 10 percent or higher for the physiological case (8,15,21), and a coefficient of 0 percent or close to 0 percent for periodic or near-periodic, artificial stimulations. It has been suggested that the magnitude of the coefficient of variation of the ISI affects the details of the EMG spectrum (8,10,15,16,21,26). In particular, changes in the median frequency of the EMG spectrum of periodically stimulated muscle are directly dependent on the mean stimulation rates because of the dependence of the spectrum on the coefficient of ISI variation. We hypothesize that the relation between the median frequency and the mean stimulation rate may not accurately reflect the actual relation between these parameters in the physiological case. The purpose of this study was to test analytically the relation between the median frequency of the EMG spectrum and the mean stimulation rate for varying coefficients of variation of the ISI, and for varying shapes of the compound motor unit action potentials (CMUAP, from MUAP, motor unit action potentials).

Based on findings from electrophysiological studies of skeletal muscle, we have used a distributed, random ENMS scheme that allowed for controlling the recruitment of motor units and the ISI statistics. The power spectrum of the EMG signal, commonly used in studies of voluntary muscular contractions (10,20,23,27,28), was chosen as an indicator of how well the EMG produced using the ENMS corresponded to that of isometric voluntary contractions. Since it is difficult to evaluate each particular ISI modification experimentally, a mathematical model, similar to that of myoelectric signal generation (8,10,17), was developed to predict the EMG spectrum of a muscle for a given stimulation design. The median frequency of the EMG power spectrum was calculated, and its dependence on the ISI statistics and the CMUAP was studied. Results of theoretical analyses confirmed experimental results reported previously using systematic, distributed, random stimulation of 8–10 ventral root filaments of cat soleus muscle (26,29).

METHOD

Advantages and Limitations of Periodic Stimulation Techniques

Periodic stimulation protocols are easy to implement, theoretically and experimentally, and they allow the study of many questions not directly concerned with simulating muscle behavior under physiologic, or near-physiologic, conditions. Periodic stimulation approaches are limited in elucidating the details of electrical, mechanical, and physiological properties of muscle during voluntary contraction, because EMG spectra (8,19), force production (29–33), rate of fatigue (unpublished observations), and many other neurophysiological features of muscle are different for periodic compared to non-periodic ENMS.

Strictly Periodic Stimulus Train

A strictly periodic stimulus train consists of a sequence of equally spaced, monophasic or biphasic, rectangular pulses. Since the duration of the stimulation pulses is much smaller than the duration of the MUAP, the stimulus train may be approximated by a sequence of impulses expressed by

$$x(t)=\sum_{j=-\infty}^{\infty}\delta(t-jT) \qquad [1]$$

where T is the reciprocal of the stimulus rate, λ :

$$\lambda=\frac{1}{T} \qquad [2]$$

The above approximation will not affect the conclusions drawn in the following analysis. The spectrum $\Phi_{xx}(f)$ of Equation 1 is obtained by

$$\Phi_{xx}(f)=\lambda^2\sum_{k=-\infty}^{\infty}\delta(f-k\lambda) \qquad [3]$$

λ which shows that the spectrum of the stimulus train is a set of delta functions with the areas of λ^2 at frequencies that are multiples of $\lambda = 1/T$, the stimulation rate. For stimulation rates within the physiological range of motor unit firing, conduction of action potentials is not inhibited; thus, the spectrum of the myoelectric response ($\Phi_{yy}(f)$) to the stimulation is

$$\Phi_{yy}(f)=\lambda^2\sum_{k=-\infty}^{\infty}\delta(f-k\lambda)\left|P(f)\right|^2 \qquad [4]$$

where P(f) is the frequency response of the system for the generation of EMG signals in a channel (i.e., the Fourier transform of the MUAP, or CMUAP, if several motor units are stimulated through a single stimulation channel). According to Equation 4, the power density spectrum (PDS) of a periodic signal is a discrete function of frequency (frequency sampling, weighted by $|P(f)|^2$). A spectrum of this type is called a line spectrum, or discrete spectrum, and it does not adequately represent EMG spectra of voluntary contractions, which have been shown to be continuous functions of frequency (1,8,32,34). Similarly, for multichannel distributed stimulations, we have

$$\Phi_{yy}(f)=\sum_{i=1}^{m}\lambda_i^2\sum_{k=-\infty}^{\infty}\delta(f-k\lambda)\left|P_i(f)\right|^2 \qquad [5]$$

where m is the number of stimulation channels, λ_i is the stimulation rate of the ith channel, and $P_i(f)$ is the Fourier transform of the CMUAP of the ith channel. As shown above for the single channel scenario, the spectrum of the multichannel periodic stimulation will also be discrete. **Figure 1** shows the theoretical spectrum of an EMG signal obtained using four stimulation channels with different rates of stimulation: 26, 30, 45, and 50 pps. As a contrast to **Figure 1**, experimental spectra of EMG signals obtained from voluntary contractions of human rectus femoris muscle are shown in **Figure 2** (2); a similar result can also be found in (19). A similar spectrum can also be obtained during voluntary contraction in other muscles. It is evident that varying rates of stimulation in different channels does

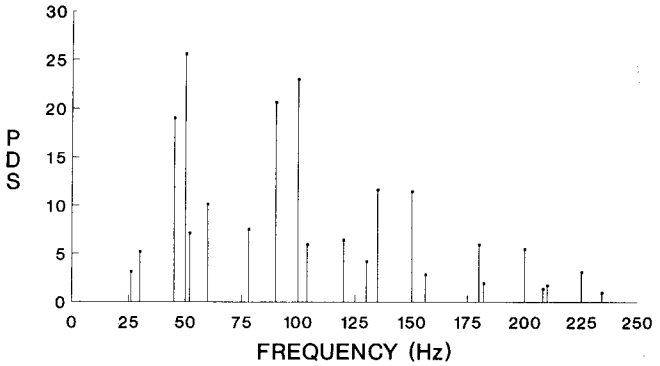

Figure 1.
Theoretical spectrum of the EMG signal obtained from four channels stimulated at periodic rates of 26, 30, 45, and 50 pps.

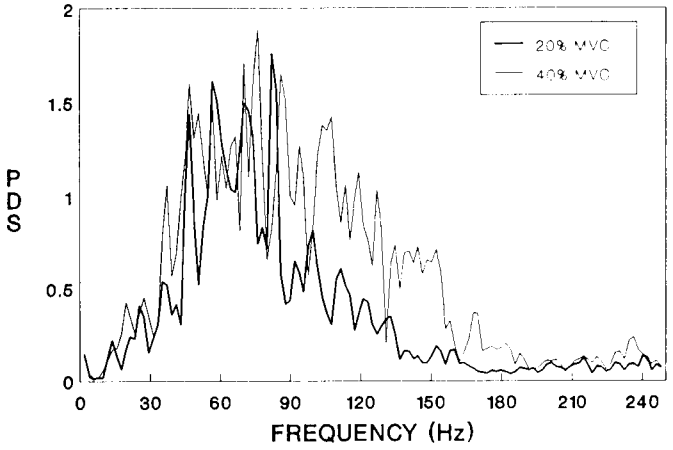

Figure 2.
Experimental spectra of EMG signals of voluntary contractions obtained from human rectus femoris at contraction levels of 20% and 40% of the maximum voluntary contraction (MVC).

not eliminate the discrete pattern of the spectrum if a periodic stimulation protocol is used. EMG spectral continuity cannot be achieved by recruiting additional motor units in a periodic ENMS protocol. However, recruitment may have an influence on the overall shape of the spectrum through the term $|P_i(f)|$ (Equation 5) which depends on the number of independent channels used for stimulation.

Practical Limitations of Periodic Stimulation

Periodic pulse trains are easy to implement, and they work well as long as physiologic properties of voluntary muscle contractions are not a major concern. However, if electrophysiological properties of muscle during voluntary contractions are the focus of study, there are at least three ways in which strictly periodic stimulation protocols are limited. First, periodic stimulations are limited in reproducing EMG spectra similar to those obtained during voluntary contractions. Stimulation-related information is carried in the temporal interval patterns of the CMUAP train. In voluntary contraction of muscle, these time interval patterns contain mainly low-frequency information (10,15,16,21,35). However, when using periodic stimulation protocols, the whole range of the EMG spectrum is affected, and thus periodic ENMS may obscure important details in the low-frequency regions of the spectrum.

Second, due to overlapping of muscle twitches and early depression effects, forces produced using periodic stimulation may differ from forces obtained using random ENMS (30,31), even for identical mean stimulation rates and recruitment strategies. As a consequence, the EMG-force relationship determined using periodic ENMS may not reflect the actual relationship that exists between these two parameters during voluntary contraction (29,32,33).

Third, the EMG spectra obtained using periodic stimulation become "discrete" (or line spectra) as shown in Equation 5. The use of this type of spectrum for determining the frequency response of muscle, or for studying systems identification, may have serious consequences, especially for high stimulation rates where only a few frequency samples are available in the EMG spectrum. This problem is enhanced when periodic stimulation is used to study vibromyographic signals or muscle sounds, which contain lower frequency components than the corresponding EMG signals (23,36–38).

Theoretical Basis For Distributed Random ENMS

The idea of using distributed random ENMS is based on findings from electrophysiological studies of skeletal muscle, specifically from a structural rather than a phenomenologic model of EMG signal generation. The structural model provides insight into how physiological parameters may contribute to observed EMG signals. It has been widely used in the areas of estimation and detection of myoelectric control (27), electrophysiological modeling (10), muscle tremor (35), performance analysis of myoelectric control channels (39–41), and generation and analysis of myoelectric signals (10,14,18). In this work, the structural model will be used to study systematically the effect of interactions between ISI patterns and CMUAPs on EMG spectra and their frequency parameters produced by distributed random ENMS. In particular, the relation between mean stimulation rate and median frequency of the EMG spectrum will be investigated as a function of varying coefficients of variation of the ISI statistics.

Spectral Expression for Distributed Multichannel Random Stimulation

A structural model for distributed ENMS is shown in **Figure 3**. In this model, $u_i(t)$, i = ,2,...,m, is the activation signal (stimulus train) which corresponds to the innervation signal for the fibers of the *i*th channel, and it is considered, as in the physiological case (10,27), a renewal point process with known activation statistics. Let $x_i(t)$, i = 1,2,...,m, represent the *i*th channel signal, then the autocorrelation function $\phi_{yy}(\tau)$ of the EMG signal y(t) can be written as

$$\Phi_{yy}(\tau) = E\left\{\sum_{i=1}^{m} x_i(t) \sum_{j=1}^{m} x_j(t+\tau)\right\} \qquad [6]$$

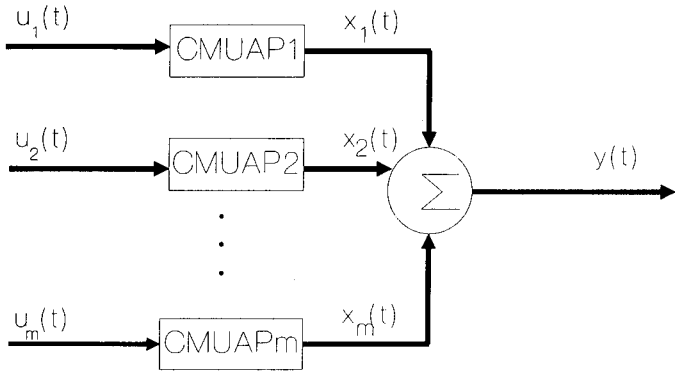

Figure 3.
Structural model of distributed ENMS with m stimulation channels.

where τ is the correlation lag value. With uncorrelated channels, the PDS, $\Phi_{yy}(f)$, is given by

$$\Phi_{yy}=(f)=\sum_{i=1}^{m}\Phi_{uui}(f)\left|P_i(f)\right|^2 \qquad [7]$$

where $\Phi_{uui}(f)$ is the PDS of the innervation process $u_i(t)$. The PDS of a single channel EMG signal consists of two parts: one part, $|P(f)|^2$, comes from the CMUAP owing to the occurrence of muscle activation, and the other part, $\Phi_{uui}(f)$, comes from the stimulation patterns of the *i*th channel. For a given channel, the PDS of the point process is found by (34,42),

$$\Phi_{uui}(f)=\lambda_i\left(1+\frac{F_{ix}(f)}{1-F_{ix}(f)}+\frac{F_{ix}^*(f)}{1-F_{ix}^*(f)}\right),\ f\neq 0 \qquad [8]$$

where the superscript * represents the complex conjugate, and $F_{ix}(f)$ is the Fourier transform of the ISI probability density function $f_{ix}(x)$. Substituting Equation 8 for $\Phi_{ii}(f)$, Equation 7 yields

$$\Phi_{yy}(f)=\sum_{i=1}^{m}\lambda_i\left(1+\frac{F_{ix}(f)}{1-F_{ix}(f)}+\frac{F_{ix}^*(f)}{1-F_{ix}^*(f)}\right)\left|P_i(f)\right|^2,\ f\neq 0 \qquad [9]$$

Equation 9 shows that 1) the EMG spectrum of simultaneous multichannel stimulation is the linear summation of the spectra for stimulation of the individual channels; and that 2) the spectrum of EMG signals does not only depend on the spectrum $P_i(f)$ of the CMUAP, but also on related physiological parameters: λ_i, the mean stimulation rate of each motor unit; m, the number of active motor units; and the pdf $f_{ix}(x)$ of the ISI.

Thus, any attempt of using the spectrum of an individual CMUAP as a measure of the entire EMG signal may, and typically will, give incomplete and misleading results. This problem occurs because the spectrum of a single CMUAP is, in general, not representative of the temporal and spatial summation of CMUAPS as a whole (Equation 9). Equation 9 allows for investigating the effect of interactions between the ISI statistics and the CMUAPs on the corresponding EMG spectra.

Spectral Properties of Random Gaussian Stimulations

Interspike intervals of voluntary contractions of skeletal muscles may be approximated using a Gaussian distribution (9). In this section, selected properties of the spectra produced by Gaussian stimulation protocols are summarized.

With $f_{ix}(x)$ representing a Gaussian distribution, Equation 9 may be written as,

$$\Phi_{yy}(f)=\sum_{i=1}^{m}\lambda_i\frac{\sinh[2(\pi f\sigma_{ix})^2]}{\cosh[2(\pi f\sigma_{ix})^2]-\cos(2\pi f/\lambda_i)},\ f\neq 0 \qquad [10]$$

where σ_{ix} is the standard deviation (SD) of the ISI for the *i*th channel. **Figures 4** and **5** show results obtained using Equation 10, for m = 1 and with different stimulation rates and coefficients of variation ($c_i=\sigma_{ix}\lambda_i$) of the ISI. In order to compare the spectral shifts for different values of the mean stimulation rate, for different values of the SD of the ISI, and for different coefficients of the ISI variation, the amplitudes of the PDS in **Figures 4** and **5** were normalized with respect to their maxima. Inspection of these figures and the corresponding equations indicates the following features:

1. The PDS, $\Phi_{uu}(f)$, for Gaussian point processes, has a high-pass characteristic. The band-pass width is controlled by the stimulation rate, λ, and the coefficient of variation of the ISI, *c*. This can be seen by the shift of the EMG spectrum as a function of the stimulation rate (**Figure 5**). At higher frequencies, the PDS, $\Phi_{uu}(f)$, is almost equal to the mean stimulation rate, λ (the normalized spectrum approaches 1 as the frequency increases). This result may be derived from Equation 10, for $f\rightarrow\infty$,

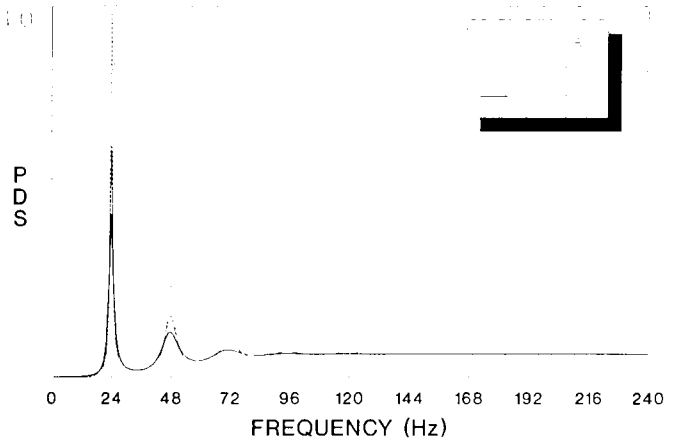

Figure 4.
Power density spectra of a Gaussian point process with $\lambda = 24pps$ and different values of c, where $c = \sigma_x\lambda$.

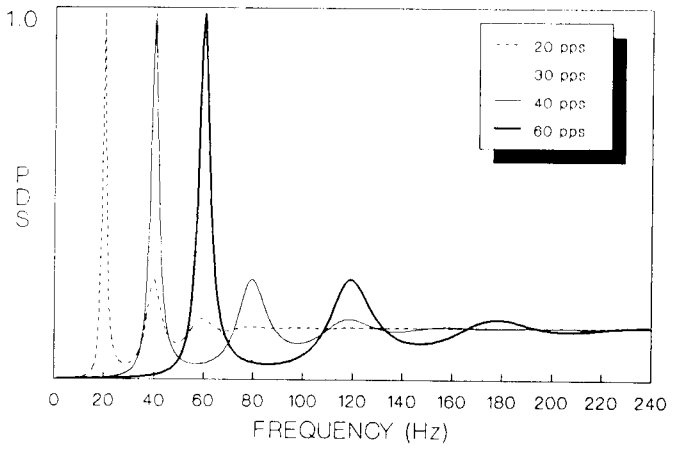

Figure 5.
Power density spectra of a Gaussian point process with $c = 0.1$ and different values of λ.

$$\lim_{f\to\infty}\Phi_{uu}(f)=\sum_{i=1}^{m}\lambda_i=\lambda_p, for\ \sigma_{ix}=0 \qquad [11]$$

where λ_p is a pooled stimulation rate.

2. The PDS, $\Phi uu(f)$, has peaks at harmonics of the firing rate, depending on the form of the ISI pdf $f_x(x)$. This result may be inferred from Equation 10. With $f = k\lambda_i, k = 1,2,...$, the cosine factor in Equation 10 produces the peaks in the PDS.

3. The magnitudes of the peaks of the PDS that are caused by the mean rate of stimulation depend on the coefficient of variation of the ISI. This may be illustrated by substituting $f = k\lambda_i$ into Equation 10 as follows:

$$\Phi_{uu}(k,c)=\sum_{i=1}^{m}\lambda_i\frac{\sinh[2(\pi kc_i)^2]}{\cosh[2(\pi kc_i)^2]-1}, f\neq 0 \qquad [12]$$

When c_i is small, the peaks of the PDS become large. The number of distinct peaks depends on the value of c_i. This statement may be verified using Equation 12, and it is illustrated in **Figure 4**. In the physiologic case, peaks of the PDS are pronounced in a frequency range from 0–120 Hz, depending on the value of c_i, and the shape of the CMUAP. This observation implies that the effect of ISI statistics on the power density spectrum of the CMUAP is mainly a low-frequency effect as reported in the literature (15,21).

4. The local minima in the PDS are given, at $f = k\lambda_i/2$ (k = 3,5,7,...), by

$$\Phi_{uu}(k,c)=\sum_{i=1}^{m}\lambda_i\frac{\sinh[(\pi kc_i)^2/2]}{\cosh[(\pi kc_i)^2/2]+1}, f\neq 0 \qquad [13]$$

The magnitudes of these minima will change as a function of c_i and k as speculated by Christakos (35).

5. The PDS becomes discrete (line spectrum) as the coefficient of variation of the ISI approaches zero, $\sigma_{ix}\to 0$. Taking the limit $\sigma_{ix}\to 0$, Equation 10 gives

$$\lim_{\sigma_{ix}\to 0}\Phi_{uu}(f)=\infty, f=\pm k\lambda_i, k=1,2,... \qquad [14]$$
$$\begin{cases} \\ 0, \text{ other values of } f\neq\pm k\lambda_i, k=0,1,2,... \end{cases}$$

When $\sigma_{ix} = 0$, (i = 1,2,...m) or $c_i = 0$, the signal becomes periodic, and a periodic signal always has a line spectrum as shown above.

6. With $\sigma_{ix} = \sigma$ and $\lambda_{ix} = \lambda$, Equation 10 becomes

$$\Phi_{uu}(f)=\lambda_p\frac{\sinh[2(\pi f\sigma)^2]}{\cosh[2(\pi f\sigma)^2]-1}, f\neq 0 \qquad [15]$$

This result shows that the spectrum of a stimulation protocol involving m channels with the same mean stimulation rate λ (but not necessarily identical stimulation patterns), and the same SD σ, will be equal to m times the spectrum of a stimulation protocol involving a single channel with λ and σ.

A Model for CMUAP

If several motor units are stimulated simultaneously in a ventral root filament, the resultant CMUAP, p(t), is the sum of the action potentials of the individual motor units in the filament. The CMUAP of a given filament will be invariant in form because of the fixed number of motor units activated, and the 'all-or-none' nature of action potential generation in motor units. The p(t) will not be identical to the muscle fiber action potential owing to diameter, endplate, and threshold dispersion, and it may not be identical across filaments for the same reasons, and because of the variation in the number of motor units in an activated filament. However, in general, the CMUAP may be expressed by summing the MUAPs activated in a single channel. One expression of a CMUAP that agrees well with observed data, and for which the spectrum can be derived analytically, is as follows:

$$p(t) = \sum_{n=1}^{L} h_n(t - \tau_n) \qquad [16]$$

where τ_n is a time shift ($\tau_1 = 0$), L is the number of motor units in a single channel, and h_n is the nth motor unit action potential in the filament which can be modelled by (27)

$$h_n(t) = \begin{cases} t(2 - b_n t)\exp(-b_n t), & 0 \le t \\ 0, & otherwise. \end{cases} \qquad [17]$$

In this Equation, b_n is a constant, or a shape factor, which is determined by the size of motor units and the distribution of fiber types in a muscle. It can be shown that the spectrum of this CMUAP is characterized by the bandwidth control parameter b_n, and the time shift τ_n:

$$P(f) = \sum_{n=1}^{L} H_n(f)\exp(-j2\pi f \tau_n) \qquad [18]$$

where $H_n(f)$ is the Fourier transform of the MUAP $h_{n(t)}$ and is given by

$$H_n = \frac{j2\pi f}{(j2\pi f - b_n)^3} \qquad [19]$$

By properly selecting the parameters b_n, and n = 1,2,...L, a spectrum similar to that of a real CMUAP can be derived. In the ideal case, one stimulation channel should only include a single motor unit. In such a situation, $\tau_1 = 0$, $P(f) = H(f)$ and the amplitude of the spectrum may be obtained using

$$|P(f)|^2 = \frac{(2\pi f)^2}{[(2\pi f)^2 + b^2]^3} \qquad [20]$$

Spectra from Equation 20 with different values of the parameter b are shown in **Figure 6**. The spectrum shifts toward higher frequencies with increasing values of b.

Effects Of Interaction Between CMUAP and ISI Statistics On Frequency Parameters

Effects on the EMG Spectrum

After having examined the properties of the ISI statistics and the model for the CMUAP, it is possible to study the effects of interactions between the ISI statistics and the CMUAP on the PDS of the EMG signal. Equation 21 shows the EMG spectrum of an ENMS with a Gaussian ISI.

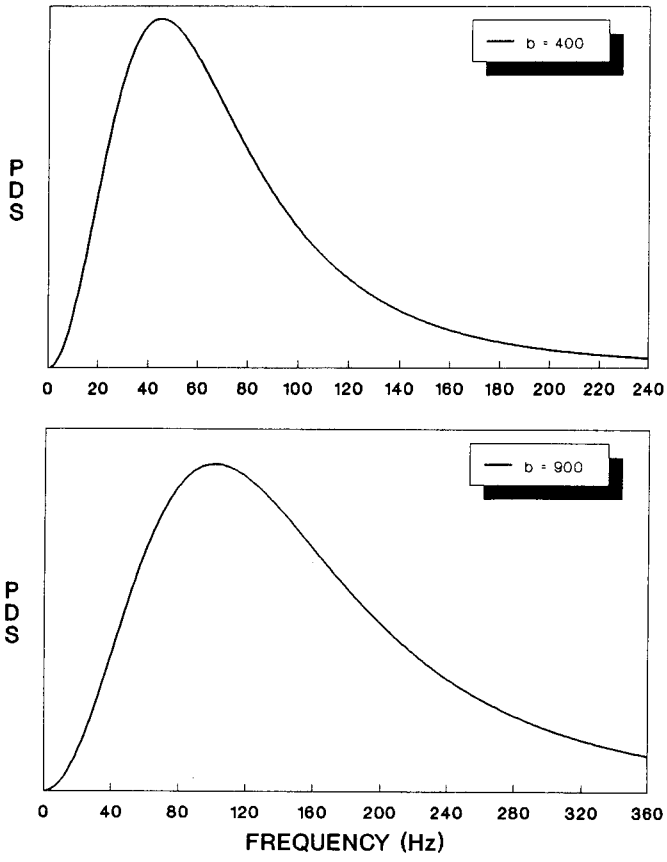

Figure 6.
Power density spectra of a MUAP with values of the parameter of $b = 400$ and $b = 900$.

$$\Phi_{yy}(f) =$$

$$\sum_{i=1}^{m} \frac{\lambda_i (2\pi f)^2 \sinh[2(\pi f \sigma_{xi})^2]}{[\cosh[2(\pi f \sigma_{xi})^2] - \cos(2\pi f/\lambda_i)][(2\pi f)^2 + b_i^2]^3}$$

$$f \neq 0 \qquad [21]$$

The EMG spectrum in Equation 21 is plotted in **Figures 7a, 7b,** and **7c** for a CMUAP with $b = 900$ and in **Figure 8** for a CMUAP with $b = 400$, for different values of λ and σ_x. These figures illustrate the changes in the EMG spectrum associated with the ISI statistics under various conditions. The influence of the ISI statistics on the PDS is mainly concentrated around the low-frequency region when σ is small (**Figures 7a** and **8**). Peaks are introduced at multiples of the firing rate, as observed experimentally in electrophysiological studies (16,21). Such changes in the PDS as a function of the ISI statistics may reflect variations in activation. For example, when activation levels increase, it is expected that the firing rate will increase, and as a result, the peak of the spectrum will shift toward higher frequencies. It is also noted from these figures, and the corresponding equations, that the magnitude of the peaks of the PDS depends on the coefficient of the ISI variation.

With a Gaussian ISI, $\sigma_{ix} = \sigma$ and $\lambda_{ix} = \lambda$, Equation 9 becomes

$$\Phi_{yy}(f) = \lambda \frac{\sinh[2(\pi f \sigma)^2]}{\cosh[2(\pi f \sigma)^2] - 1} \sum_{i=1}^{m} |P_i(f)|^2, \quad f \neq 0 \quad [22]$$

Equation 22 gives the PDS of a distributed, random ENMS, where stimulations in each channel have similar rather than identical statistical properties. When the CMUAPs are approximately the same across channels, the PDS of the EMG signal has the same form as that of an individual channel, except that the multichannel PDS is scaled by a factor m, where m is the number of channels used in the ENMS. This result represents the so called grouping effect of action potentials and has been reported in electrophysiological studies (35).

Effects on the Median Frequency

In order to quantify changes in the myoelectric spectrum as a function of the firing rate and SD of the interspike interval, the median frequency, f_{med}, of the signal was calculated numerically using Equation 21 (with m = 1) as a function of λ. The plots of the median frequency against λ are shown in **Figure 9** for different values of σ_x and in **Figure 10** for different values of the parameter b.

The relationship between the median frequency of the EMG spectrum and λ is nonlinear, especially for small σ_x. This result is associated with the nonlinear relation between $\Phi_{yy}(f)$ and λ, as indicated in Equation 21, and in

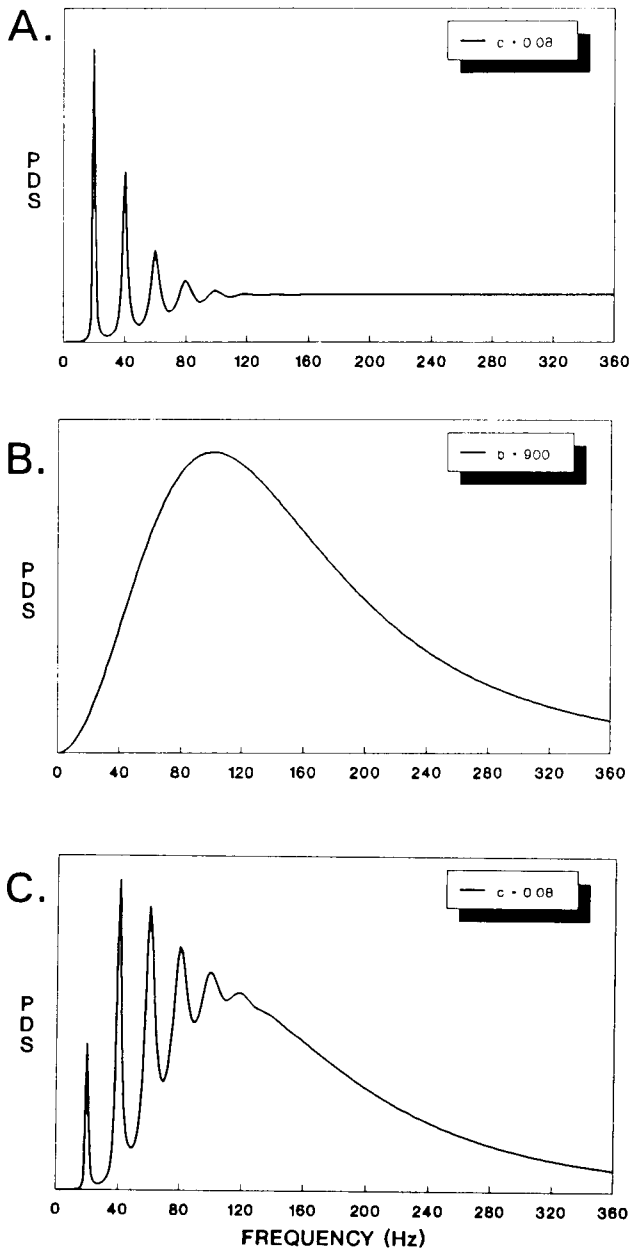

Figure 7.
Influence of the ISI statistics on the PDS of the EMG signal: (a) PDS of a Gaussian point process with $\lambda = 20pps$; (b) MUAP with parameter $b = 900$; and, (c) PDS of the EMG signal.

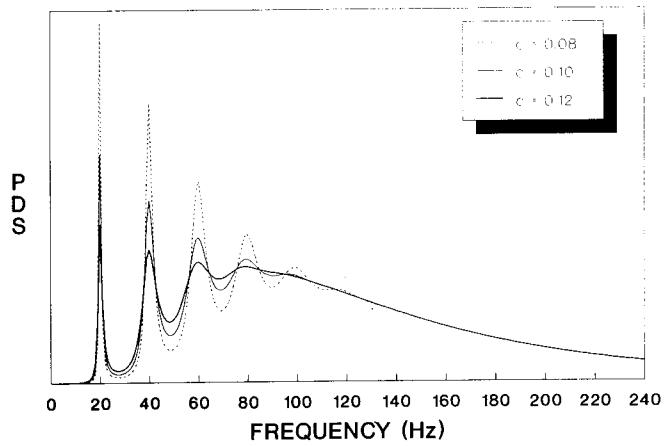

Figure 8.
Effects of changes in the coefficient of variation of the ISI on the PDS of the EMG signal with $\lambda = 20pps$.

Figures 8 and **9**. For large values of σ_x, the change in the median frequency as a function of λ becomes small (e.g., **Figure 9**, $\sigma_x = 8$ms), because the fact that the effect of changing λ on the EMG spectrum decreases with increasing σ_x. Nevertheless, in all cases, the median frequency tends to increase with increasing λ. These data support experimental results reported in the literature (21).

The observation that changes in the median frequency, as a function of λ, become small for large values of the σ_x is further demonstrated theoretically in the Appendix using m = 1. It was found that for large values of the SD of the ISI, the median frequency relates to parameter b of the MUAP as,

$$f_{med} \leq 5|b| \qquad [23]$$

Equation 23 indicates that the median frequency is independent of the rate of stimulation (or firing), when the values of the σ_x are large, but depends on the CMUAP through the parameter b. The larger the parameter b, the higher the median frequency. In general, the median frequency depends on both stimulation (or firing) statistics, and the MUAP, and thus is an indicator of the combined effects of nerve stimulation and muscle response to the nerve stimulation.

Experiments on Cat Soleus Muscle

Experimental data were obtained from the cat soleus muscle. Cats were anesthetized and placed in a stereotaxic frame (43) in a prone position with the hind limbs rigidly fixed. Ventral roots L7 and S1 were exposed, separated, and carefully divided into bundles (29). Each of these bundles was hung over a separate bipolar electrode for individual and simultaneous distributed stimulations. The stimulations, using patterns generated from computer simulations, were applied to the ventral root bundles via the electrode. EMG signals from stimulations of each of the 10 nerve bundles individually (finger prints), and the simultaneous stimulation of all 10 bundles were measured using a pair of indwelling bipolar electrodes. EMG signals were digitized on line and stored on a computer. Blood

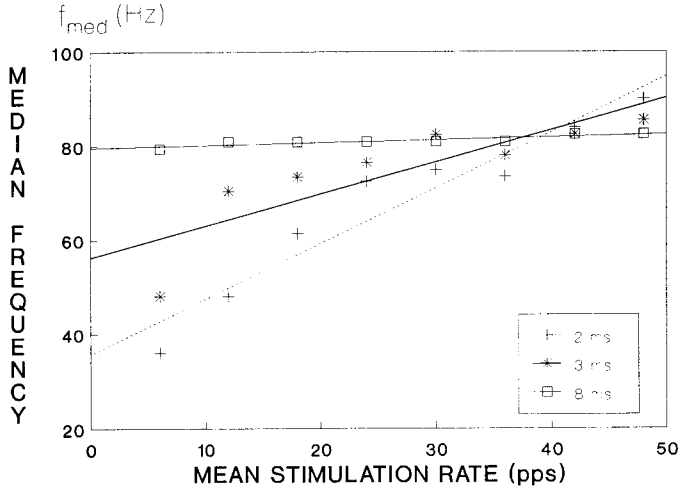

Figure 9.
Median frequency of PDS versus λ for $\sigma_x = 2$, 3 and 8 ms. (Symbols: the calculated median frequency, Lines: the regression line approximation).

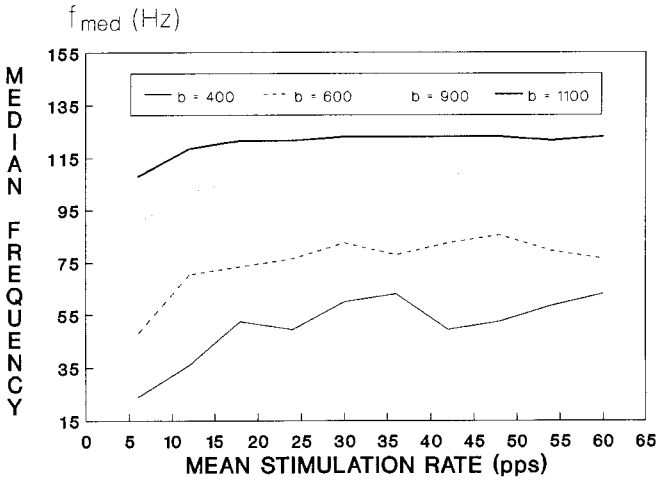

Figure 10.
Median frequency of the PDS as a function of λ for different values of the MUAP parameter b.

pressure, core temperature, and muscle temperature of the cats were monitored continuously and were kept constant throughout the experiment.

The power density spectrum of each experimental EMG signal was estimated using a 512-point fast Fourier transform (FFT) algorithm, and then averaged over four consecutive segments in order to reduce estimation errors. **Figure 11** shows two representative spectra of experimental EMG signals obtained using the same value for μ and different values for *c*. The experimental results support the mathematical predictions discussed in the previous sections. Specifically, the EMG signal spectrum shows peaks at the mean stimulus rate and its multiples when the coefficient of the ISI variation is small; and the envelope of the PDS is determined by the CMUAP waveform and the details of the ISI stimulus statistics.

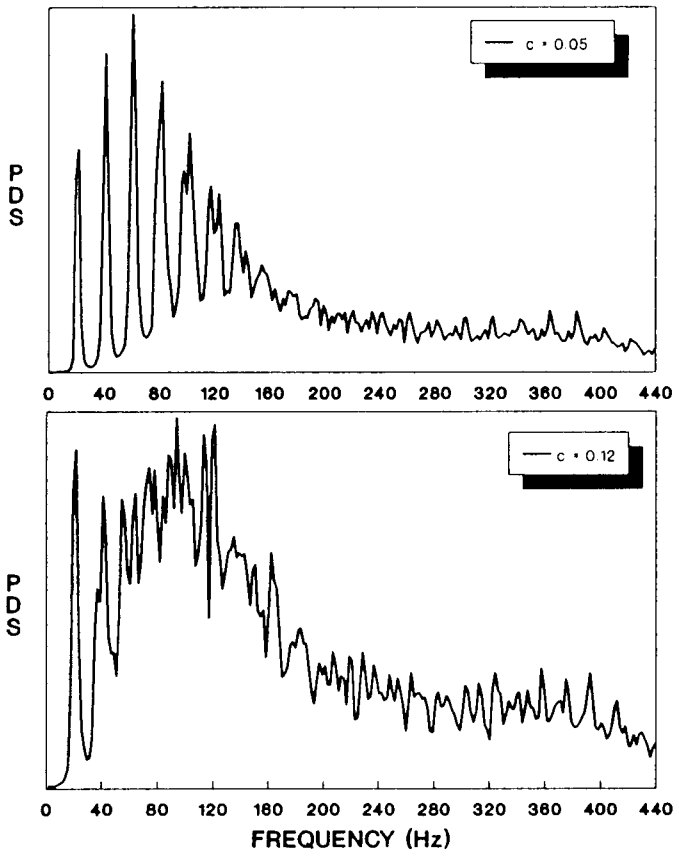

Figure 11.
PDS of EMG signals obtained from random ENMS with $\lambda = 20pps$ and different values of *c*.

RESULTS AND DISCUSSION

General Discussion

Figure 8 shows the effects of changes in the SD of the ISI on the single channel EMG spectrum. A decrease in the SD of the ISI produced an increasingly more discrete spectrum of the EMG, indicating that variations of the ISI play an important role in controlling the details of the corresponding EMG spectrum. This finding was supported by our experimental results (**Figure 11**). A comparison of the experimental results of the ENMS (**Figure 11**) with the spectra obtained during voluntary contraction in the human rectus femoris (**Figure 2**) indicates that the EMG spectrum obtained using ENMS with a coefficient of variation of the ISI of 12 percent is similar to that obtained for voluntary contractions. Therefore, in situations where it is necessary to produce EMG spectra and frequency parameters similar to those obtained during voluntary contractions, it is suggested that variations in the inter-stimulus intervals be introduced.

The theoretical results of this study indicate 1) that the EMG spectrum shows peaks at the mean stimulation rate and its multiples; 2) that the magnitude, or clarity, of the peaks at the mean stimulation rate and its multiples depends on the coefficient of variation of the ISI, and further, that the PDS approaches a line spectrum when the coefficient of variation of the ISI becomes small (**Figure 8**); and, 3) that the envelope of the PDS is primarily determined by the shape of the CMUAP, and is virtually independent of the ISI statistics when the coefficient of variation of the ISI is large. This result is illustrated by the similarity of the envelopes of the spectra in **Figures 7** and **8**. It has been found that, from a mathematical point of view, there is a distinct difference in EMG spectra between periodic and random ENMS. It is possible to obtain "normal" EMG spectra (as defined here) with random ENMS, but not with periodic ENMS.

The median frequency of the PDS depends on the interaction of the firing (or stimulation) statistics and the MUAP. The results reported here imply that it is possible to use the median frequency as a selective indicator of changes in the MUAP by increasing the SD of the inter-stimulus interval in distributed random ENMS to sufficiently large values (8 ms or higher, see **Figure 9**). This approach may be used when attempting to determine qualitatively the contribution of variations in CMUAP to a change of the median frequency using random ENMS.

In summary, three features distinguish the present study from previous ENMS investigations (6). First, emphasis was placed on introducing random ISI into the ENMS. Second, the distributed multichannel stimulation

approach allows the simulation of recruitment of motor units according to the size principle. This is an important feature of physiologic activation. Third, the effects of the mean stimulation rate on the EMG spectrum and the frequency parameters of stimulated muscles were studied analytically for varying coefficients of variation of the ISI and for varying shapes of the CMUAP. The theoretical findings presented in this study are supported by the experimental findings using a distributed, random ENMS approach.

Clinical Relevance

The median frequency of the EMG spectrum has been used widely to assess muscular fatigue. When muscles fatigue, the median frequency of the EMG signal tends to decrease (8). However, in experiments using voluntary muscular contraction, it is difficult to determine how much of the change in the median frequency is caused by changing in the local muscle properties (as reflected in the MUAP) and how much is caused by the decrease in firing rates of motor units. Distributed random ENMS provides a possibility for separating these two factors that influence the median frequency by choosing the SD of the ISI appropriately.

CONCLUSION

The distributed, random ENMS approach described here can be used to study the properties of EMG spectra in situations approximating voluntary contractions; however, simply reproducing EMG spectrum of voluntary contraction is not sufficient to guarantee mechanical muscle behavior similar to that of voluntary contractions. The effect of changes in stimulation parameters, such as the mean stimulation rate, the number of stimulation channels, and the coefficient of variation of the ISI, on muscle force, EMG, and VMG signals must be determined analytically and experimentally.

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REFERENCES

1. Mannard A, Stein RB. Determination of the frequency response of isometric soleus muscle in the cat using random nerve stimulation. J Physiol 1973:229:275–84.
2. Petrofsky JS. Control of the recruitment and firing frequencies of motor units in electrically stimulated muscles in the cat. Med Biol Eng Comput 1978:16:302–8.
3. Rack P, Westbury D. The effect of length and stimulus rate on tension in the isometric cat soleus muscle. J Physiol 1964:204:443–60.
4. Solomonow M. External control of the neuromuscular system. IEEE Trans Biomed Eng 1984:31:752–63.
5. Solomonow M, Baratta R, Zhou BN, Shoji H, D'ambrosia RD. The EMG-force model of electrically stimulated muscle: dependence on control strategy and predominant fiber composition. IEEE Trans Biomed Eng 1987:34:692–703.
6. Stokes MJ, Cooper, RJ. Muscle sounds during voluntary and stimulated contractions of the human adductor pollicis muscle. J Appl Physiol 1990:72:1908.
7. Zhang YT, Hulliger M, Herzog W. The EMG spectrum of electrically stimulated muscle: dependence on the interspike interval statistics. In: Proceedings of the 17th CMBE Conference. Banff, Canada, 1991:107–8.
8. Basmajian J, DeLuca CJ. Muscles alive. Baltimore: Williams and Wilkins, 1985.
9. Clamann HP. Statistical analysis of motor unit firing patterns in a human skeletal muscle. Biophys J 1969:9:1233–51.
10. DeLuca CJ. Physiology and mathematics of myoelectric signals. IEEE Trans Biomed Eng 1979:26:313–25.
11. DeLuca CJ, LeFever R, McCue M, Xenakis A. Behavior of human motor units in different muscles during linearly varying contraction. J Physiol 1982:329:113–28.
12. Milner-Brown HS, Stein RB, Yemm R. Changes in firing rate of human motor units during linearly changing voluntary contractions. J Physiol 1973:230:371–90.
13. Milner-Brown HS, Stein RB, Yemm R. The orderly recruitment of human motor units during voluntary isometric contractions. J Physiol 1973:228:285–306.
14. Brody G, Scott RN, Balasubramanian R. A model for myoelectric signal generation. Med Biol Eng 1974:12:29–41.
15. Lago P, Jones NB. Effect of motor-unit firing time statistics on EMG spectra. Med Biol Eng Comput 1977:15:648–55.
16. Paiss O, Inbar GF. Autoregressive modeling of surface EMG and its spectrum with application to fatigue. IEEE Trans Biomed Eng 1987:34:761–70.
17. Zhang YT, Parker PA, Scott RN. A study of the effects of motor unit recruitment and firing statistics on the signal to noise ratio of myoelectric control channel. Med Bio Eng Comp 1990:28:225–31.
18. Graupe D, Salahi J, Zhang DS. Stochastic analysis of myoelectric temporal signatures for multifunctional signal-site activation of prostheses and orthoses. J Biomed Eng 1985:7:18–29.
19. Hagg GM. Interpretation of EMG spectral alterations and alteration indexes at sustained contraction. J Appl Physiol 1992:73:1211–7.
20. Lindstrom LH, Magnusson RI. Interpretation of myoelectric power spectra: a model and its applications. Proc IEEE 1977:65:653–62.

21. Pan ZS, Zhang YT, Parker PA. Motor unit power spectrum and firing rate. Med Biol Eng Comput 1989:27:14–8.
22. Zhang YT, Parker PA, Scott RN. Modifications of ME signal power density spectrum. In: Proceedings of the IEEE Engineering in Medicine and Biolology Society's 12th Annual International Conference. 1990:12:2198–9.
23. Zhang YT, Frank CB, Rangayyan RM, Bell GD. A comparative study of vibromyography and electromyography obtained simultaneously from active human quadriceps. IEEE Trans Biomed Eng 1992:39:1045–52.
24. Zhou BH, Baratta R, Solomonow M. Manipulation of muscle force with various firing rate and recruitment control strategies. IEEE Trans Biomed Eng 1987:34:128–39.
25. Baratta R, Zhou BH, Solomonow M. Frequency response model of skeletal muscle: effect of perturbation level, and control strategy. Med Biol Eng Comput 1989:27:337–45.
26. Zhang YT, Parker PA, Herzog W, Hulliger M, Guimaraes A. Distributed random electrical neuromuscular stimulation: effects of inter-stimulus interval statistics on EMG spectrum. In: Proceedings of the RESNA International Conference, 1992, Toronto, ON. Washington, DC: RESNA Press, 1992:235–7.
27. Parker PA, Scott RN. Myoelectric control of prostheses. CRC Crit Rev Biomed Eng 1986:13:284–310.
28. Zhang YT, Parker PA, Scott RN. Low-pass effects of propagation delay dispersion in single motor unit. In: Proceedings of the IASTED International Symposium on Computers and Advanced Technology in Medicine, Healthcare and Bioengineering. Honolulu, HA 1990:40–1.
29. Guimaraes A, Herzog W, Hulliger M, Zhang, YT, Day S, EMG-force relation of the cat soleus muscle studied with distributed and non-periodic stimulation of ventral root filaments. J Exp Bio 1994:March.
30. Astrand PO, Rodahi, K. Textbook of work physiology: physiological bases of exercise. 2nd ed. New York: Mcgraw-Hill. 1977.
31. Burke RE, Edgerton VR. Motor unit properties and selective invovement in movement. Exer Sport Sci Rev 1975:3:31.
32. Guimaraes A, Herzog W, Zhang YT, Hulliger M, Day S. EMG-force relationship of mammalian skeletal muscle (part I): rate modulation of a ventral root filament containing several motor units. In: Proceedings of NACOB II: The Second North American Congress on Biomechanics, 1992:175–6.
33. Guimaraes A, Herzog W, Hulliger M, Zhang YT, Day S. EMG-force relationship of mammalian skeletal muscle (part II): experimental simulation of recruitment and rate modulation using ten ventral root filaments. In: Proceedings of NACOB II: The Second North American Congress on Biomechanics, 1992:193–4.
34. Cox DR, Lewies PAW. The statistical analysis of series of events. London: Methuen, 1966.
35. Christakos CN. A study of the electromyogram using a population stochastic model of skeletal muscle. Biol Cybern 1982:45:5–12.
36. Barry DT, Leonard JA, Gitter AJ, Ball RD. Acoustic signals from frog skeletal muscle. Biophys J 1987:51:769–73.
37. Orizio C, Perini R, Diemont B, Figini MM, Veicsteinas A. Spectral analysis of muscular sound during isometric contraction of biceps brachii. J Appl Physiol 1990:68:508–12.
38. Oster G, Jaffe JS. Low-frequency sounds from sustained contraction of human skeletal muscle. Biophys J 1980:30:119–28.
39. Zhang YT, Parker PA, Scott RN. A correlated multi-myoelectric channel model and its SNR characteristics. In: Proceedings of the IEEE Engineering in Medicine and Biolology Society's 11th Annual International Conference, 1989, Seattle, WA. 1989:1133–4.
40. Zhang YT, Parker PA, Scott RN. Effects of electrode configuration and spacing on the SNR characteristics of ME signal. In: Proceedings of the 16th CMBEC. Winnipeg, Canada: 1990:77–8.
41. Zhang YT, Parker PA, Scott RN. Control performance characteristics of myoelectric signal with additive interference. Med Biol Eng Comp 1991:29:84–8.
42. Cox DR, Smith NL. On the superposition of renewal processes. Biometrica 1954:41:91–9.
43. Herzog W, Lepnard TR, Renaud JM, Wallace J, Chaki G, Bornemisza S. Force-length properties and functional demands of cat gastrocnemius, soleus and plantaris muscles. J Biomech 1992:25:1329–35.

APPENDIX

LIST OF SYMBOLS

A—the average amplitude of the MUAP
b—the shape factor of motor unit action potential
c—the coefficient of variation $c = \sigma_x\lambda$
CMUAP—the compound motor unit action potential
$E[\ ]$—the expectation operator
ENMS—electrical neuromuscular stimulation
$f_x(x)$—the probability density function of random inter-stimulus interval x
$F_x(f)$—the Fourier transformation of $f_x(x)$
$h(t)$—the CMUAP
ISI—the inter-stimulus interval (or the inter-spike interval)
m—the number of stimulation channels
MUAP—the motor unit action potential
pdf—probability density function
PDS—power density spectrum
$P(f)$—the Fourier transform of $p(t)$
pps—pulses per second
$p(t)$—the averaged MUAP
$r(t)$—the renewal point process
SD—standard deviation
T_i—the excitation time instant
T_o—the duration of the MUAP
$var[\ ]$—the variance operator
x_i—the ith ISI
λ—the mean stimulus (firing) rate
σ_x—the standard deviation of the ISI
$\phi(\tau)$—the autocorrelation function
$\Phi(f)$—the power density spectrum

The Relation between Median Frequency and the Parameter *b*

When σ is large, we have from Equation 22 the approximation

$$\Phi_{yy}(f) \cong \sum_{i=1}^{m} A_i\lambda_i \frac{(2\pi f)^2}{\left\{(2\pi f)^2 + b_i^2\right\}^3} \qquad [A1]$$

According to the definition of the median frequency, we get

$$\int_0^{f_{med}} \sum_{i=1}^{m} A_i\lambda_i \frac{(2\pi f)^2}{\left\{(2\pi f)^2 + b_i^2\right\}^3}\, df = \int_{f_{med}}^{\infty} \sum_{i=1}^{m} A_i\lambda_i \frac{(2\pi f)^2}{\left\{(2\pi f)^2 + b_i^2\right\}^3}\, df \qquad [A2]$$

For m = 1, the above equation reduces to

$$\int_0^{f_{med}} \frac{(2\pi f)^2}{\left\{(2\pi f)^2 + b^2\right\}^3}\, df = \int_{f_{med}}^{\infty} \frac{(2\pi f)^2}{\left\{(2\pi f)^2 + b^2\right\}^3}\, df \qquad [A3]$$

Solving the definite integrals on both sides (with $x = 2\pi f_{med}$ and $e = b^2$), we get

$$\frac{x}{\pi\left[(x/2\pi)^2 + e\right]^2} + \frac{x}{2\pi e\left[(x/2\pi)^2 + e\right]} + \frac{1}{e^{1/2}} arctg\left(\frac{x}{2\pi e^{1/2}}\right) - \frac{\pi}{4e^{1/2}} = 0 \qquad [A4]$$

For $x < 2\pi e^{1/2}$, we have

$$artg\left(\frac{x}{2\pi e^{1/2}}\right) = \frac{x}{2\pi e^{1/2}} - \frac{x^3}{3(2\pi e^{1/2})^3} + ... \qquad [A5]$$

Using the first order approximation in Equation (A4) and considering $e = b^2$, we obtain

$$\frac{x^5}{2(2\pi b)^4} - \frac{x^4}{4b^3 2^4 \pi^2} + \frac{3x^3}{2b^2(2\pi)^2} - \frac{x^2}{8b} + 2x - \pi^2 b = 0 \qquad [A6]$$

Substituting $f_{med} = x/2\pi$ into the above Equation (A6), we finally get

$$\left(\frac{f_{med}}{b}\right)^5 - \frac{\pi}{\pi}\left(\frac{f_{med}}{b}\right)^4 + 3\left(\frac{f_{med}}{b}\right)^3 - \frac{\pi}{2}\left(\frac{f_{med}}{b}\right)^2 + 4\left(\frac{f_{med}}{b}\right) - \pi = 0 \qquad [A7]$$

The upper bound of the root Υ for the general equation

$$a_0 x^n + a_1 x^{n-1} + ... + a_{n-1} x + a_n = 0 \qquad [A8]$$

is given by

$$|\Upsilon| \le 1 + \frac{1}{a_0} \max\left\{|a_1|, |a_2|, ..., |a_n|\right\} \qquad [A9]$$

and

$$f_{med} \le 5\,|b| \qquad [A10]$$

A more specific root of Equation A10 could be obtained using the Newton Integration approach. For our purpose, Equations A7 and A10 are sufficient to indicate that the median frequency is independent of the stimulation (or firing rate) for large values of the standard deviation of ISI but depends on the CMUAP, which supports the observations made in Figure 6 and Equation 21.

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A Technical Note

Initial clinical evaluation of a wheelchair ergometer for diagnostic exercise testing: A technical note

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Abstract—The purpose of this initial study was to evaluate a new wheelchair ergometer (WCE) and exercise test protocol for the detection of coronary artery disease in men with lower limb disabilities. Forty-nine patients (63 ± 9 yr) completed WCE tests without complications. Peak heart rate was 84 ± 15% (mean ± SD) of age-predicted maximum and peak double product was $223 \pm 62 \times 10^2$. The specified target heart rate (≥ 80% age-predicted maximal) or a positive result was achieved in 76% of tests. Fourteen tests were rated positive, 21 as negative and 14 as nondiagnostic for exercise-induced ischemia. In 18 patients who underwent coronary angiography, the predictive value was 100% (10/10) for a positive, and 50% (2/4) for a negative WCE test result. These results suggest that WCE is a viable initial diagnostic option for some persons who cannot adequately perform treadmill or cycle ergometry exercise.

Key words: *coronary artery disease, exercise testing, exertion, upper body exercise, wheelchair ergometry.*

INTRODUCTION

Signs and/or symptoms of clinically significant coronary artery disease (CAD) may be absent in the resting state. Therefore, graded exercise tests are frequently used to induce diagnostic myocardial ischemia in cases of suspected CAD. Available evidence suggests that persons with a variety of lower limb disabilities possess a greater than average risk of CAD morbidity and mortality (1–5). However, the inability of these patients to complete exercise tests on a treadmill or cycle ergometer presents the clinician with a unique diagnostic problem.

Current clinical practice is to utilize pharmacologic modalities (e.g., dobutamine, adenosine) in combination with echocardiography or thallium scintigraphy for diagnosis of CAD in persons with lower limb disabilities. These procedures are more costly than standard treadmill or cycle exercise tests, require intravascular access, and are less convenient to perform. Diagnostic testing utilizing upper body exercise represents a possible alternative. Several studies have evaluated arm crank ergometry for this purpose (6–20), but, to the authors' knowledge, no comparable evaluations have been carried out utilizing wheelchair ergometry (WCE). Manual wheelchair propulsion is the primary mode of mobility for many persons with lower limb disabilities. Therefore, WCE offers a more familiar and task specific mode of exercise testing for these individuals. Moreover, it is anticipated that the Americans with Disabilities Act will provide impetus for the production of commercially available wheelchair ergometers suitable for diagnostic exercise testing.

The present investigation was the initial clinical evaluation of a new wheelchair ergometer and exercise test

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protocol for persons unable to adequately perform treadmill or cycle ergometry exercise.

METHODS

Subjects

The subjects for this research were 49 consecutive male patients 63 ± 9 years of age (range 42–81) who were referred to the authors' laboratory for diagnostic WCE graded exercise testing. Forty-five patients had lower limb disabilities, one patient was visually impaired and three subjects were nondisabled. Disabilities included peripheral vascular disease of the lower extremities (n = 17), lower limb amputation (n = 7), hemiplegia (n = 6), spinal cord injury (n = 5) and other musculoskeletal or neurological disorders (n = 11). Subjects were primarily inpatients referred for cardiovascular assessment before surgery, evaluation of signs or symptoms suggesting the presence of CAD, or appraisal of antianginal or antihypertensive therapy. At the time of testing, 10 patients were taking digitalis, 31 were taking antianginal agents, and 23 were taking various antihypertensive medications.

The research protocol was approved by the Human Studies Subcommittee of Hines VA Hospital and all subjects provided written informed consent to participate in this study.

Wheelchair Ergometer Exercise Tests

Each subject performed a continuous graded exercise test on a magnetic eddy current braked WCE called the Wheelchair Aerobic Fitness Trainer (21,22). In preparation for testing, patients transferred to a Quickie GP (Motion Design Corporation, Fresno, CA) wheelchair which was loaded onto the WCE. Wheelchair tire pressure was maintained at 60–65 pounds per square inch.

Subjects were required to maintain a wheel speed of 2 mi·h^{-1}, with increments in upper body work accomplished by increasing braking resistance. Power output requirements began at 6 watts (W) in the first stage and increased by 5 to 7 W per stage. Stages were 3 minutes in length. Guidelines for terminating exercise tests were those published by the American College of Sports Medicine (23).

With a Quinton 3000B Stress Test Monitor (Quinton Instrument Company, Seattle, WA), a 12-lead ECG was recorded at rest, each minute during exercise and for at least 5 minutes into recovery. Three ECG leads were continuously monitored throughout the test. Because it was not possible to assess blood pressure during upper body exercise by auscultation, measurements were taken during brief pauses between stages as follows: the blood pressure cuff was pressurized during the final 5 seconds of each stage, the subject was told to stop wheeling, and the blood pressure was taken immediately. As soon as the diastolic pressure was determined (~ 15–20 sec) the subject began the next stage.

Ratings of perceived exertion were obtained using Borg's 15-point (6–20) graded scale (23). Ratings were taken during the last 30 seconds of every stage and at peak exercise. Prior to testing, each subject was asked to look over the rating of perceived exertion scale. An investigator then read aloud the instructions for the determination of rating of perceived exertion while the subject followed on a second copy. Ratings of angina were also obtained during the last 30 seconds of each stage using a 5-point (0–4+) scale (23).

Oxygen uptake ($\dot{V}O_2$) was determined in 23 subjects by the open circuit method. A Rudolph mask or mouthpiece equipped with a non-rebreathing valve was utilized. Expired gases were analyzed with the MMC Horizon™ System (SensorMedics Corp., Yorba Linda, CA), that was calibrated before and after each test with reference gases and room air.

Interpretation of the Exercise ECG Data

Each WCE exercise test was graded by a cardiologist as positive for ischemia, negative for ischemia, or nondiagnostic. Criteria for a positive test included 1) ≥ 0.10 mV horizontal or down sloping ST segment depression, or additional ST segment depression, persisting 80 msec after the J point, during or after exercise; 2) ≥ 0.10 mV ST segment elevation persisting 80 msec after the J point, during or after exercise, in the absence of significant Q waves; or 3) exercise induced angina (23,24).

The ECG was considered nondiagnostic in the presence of 1) left bundle branch block, 2) left ventricular hypertrophy with a strain pattern, 3) current use of digitalis (if exercise induced ST segment depression was present), or 4) failure to reach 80 percent of age-predicted maximal heart rate [APMHR = 220 – age(yr)] without ischemic ST segment changes or angina. Tests were still considered positive for the above categories if patients experienced exercise induced angina. The use of digitalis is known to increase the likelihood of a false positive exercise test. However, absence of ST segment depression in patients taking digitalis who reach adequate levels of cardiovascular stress provides strong evidence against myocardial

ischemia (25). Therefore, if a subject was free of angina and ECG ST segment displacement and reached an adequate HR (≥ 80 percent APMHR), his test was categorized as negative for ischemia.

Coronary Angiography

Selective coronary angiography, utilizing the Judkins technique, was performed before or after WCE exercise testing in 18 of the 49 subjects. The decision to proceed with angiography was made by each patient's physician and was not a required aspect of the research protocol. Multiple orthogonal views were obtained of each vessel. Coronary angiograms were visually inspected by a cardiologist and considered positive for clinically significant CAD if one or more major epicardial coronary artery or branch vessel showed a reduction of ≥ 70 percent of its luminal diameter.

Statistical Analysis

Chi-square analysis was used to determine if the distribution of WCE test results was independent of medication status and whether or not subjects completed metabolic testing. Mean differences in peak exercise responses between these groups were also compared using nonpaired Student's *t*-tests. Decisions regarding statistical significance were based upon an alpha level of 0.05. All statistical analyses were carried out using the Statview 4.0 statistical analysis package (Abacus Concepts, Calabasas, CA).

RESULTS

Physiologic Responses and Exercise Test Interpretation

A total of 49 exercise tests were administered without complications. Peak exercise cardiovascular responses are summarized in **Table 1**. Fourteen tests were rated as positive for ischemia, 21 as negative for ischemia, and 14 as nondiagnostic. Of the nondiagnostic tests, 10 were due to failure to reach 80 percent of APMHR (including one patient with complete left bundle branch block). An 80 percent threshold was considered appropriate since peak heart rate is expected to be lower for upper body as compared to lower body exercise (26). Had the more traditional 85 percent cutoff been adopted, three of the negative tests would have been classified as nondiagnostic. Four of the 10 subjects with nondiagnostic tests had significant ST segment depression but were taking digitalis. Of the 14 positive WCE tests, chest pain and significant ST segment displacement were present in two, ST segment displacement without angina was noted in five, and angina without ST segment changes appeared in seven.

Metabolic measurements were completed on 47 percent (n = 23) of the subjects. There were no statistically significant differences ($p > 0.05$) in age, peak heart rate, rate-pressure product, percent of APMHR, or $\dot{V}O_2$ between subjects completing and those not completing metabolic testing. In addition, chi-square analysis showed that the frequency of WCE test results (positive, negative,

Table 1.
Mean ± standard deviation for age, peak cardiovascular and metabolic responses, rating of perceived exertion (median) and power output grouped by wheelchair ergometry test results.

Parameter	All (n = 49)	Positive (n = 14)	Negative (n = 21)	Non-diagnostic (n = 14)
Age (yr)	63 ± 9	65 ± 5	61 ± 10	66 ± 9
Heart Rate ($b \cdot min^{-1}$)	132 ± 26	125 ± 26	149 ± 21	113 ± 20
% APMHR	84 ± 15	80 ± 15	94 ± 10	74 ± 11
Systolic BP (mmHg)	169 ± 29	162 ± 29	173 ± 30	168 ± 28
RPP ($\times 10^2$)	223 ± 62	205 ± 63	256 ± 53	193 ± 55
$VO_2(L \cdot min^{-1})$	1.13 ± 0.40	0.91 ± 0.25	1.41 ± 0.47	0.99 ± 0.19
	(n = 23)	**(n = 6)**	**(n = 9)**	**(n = 6)**
RPE (median)	18.5	18	18	19
Power Output (watts)	23 ± 9	21 ± 8	23 ± 11	23 ± 6

HR = heart rate; RPP = rate pressure product; APMHR = age-predicted maximal heart rate; $\dot{V}O_2$ = oxygen uptake; RPE = rating of perceived exertion.

nondiagnostic) was independent of whether or not metabolic testing was completed ($p > 0.05$).

Peak metabolic and cardiovascular exercise responses for the entire group and patients in the three diagnostic categories are shown in **Table 1**. The peak rate-pressure product for the entire group was $223 \pm 62 \times 10^2$ and the peak heart rate was 132 ± 26 b·min^{-1} (84 ± 15 percent APMHR). Attainment of peak heart rate was precluded in a number of the positive tests where exercise was stopped due to signs or symptoms of myocardial ischemia. If positive tests are excluded, mean percent of APMHR was 86 ± 14 and the mean rate-pressure product was $231 \pm 61 \times 10^2$. Most patients (76 percent) were able to reach 80 percent of their APMHR or had positive tests (69 percent if an 85 percent cut-off is substituted) despite the fact that 55 percent of the group were taking beta adrenergic- and/or calcium channel blocking medications.

No difference in peak $\dot{V}O_2$ was found between patients taking antianginal medications (n = 30; calcium channel blockers and/or beta blockers and/or nitrates) and those not taking any of these medications (n = 19). In addition, no significant differences were found between peak $\dot{V}O_2$ of the 9 subjects taking beta receptor antagonists and the remaining 40 patients.

Coronary Angiography

Eighteen patients underwent coronary angiography before or after exercise testing. Two patients were free of significant CAD. Single-vessel CAD was found in six patients, two-vessel disease in seven, and triple-vessel disease in three (see **Table 2**). There were ten true positive and two true negative WCE tests. Two patients had false negative tests, one subject with single-vessel disease of the circumflex artery and the other with two-vessel disease. Four patients with nondiagnostic tests had positive angiograms. It should be noted that three of these patients had significant ST segment depression but their tests were classified as nondiagnostic because they were taking digitalis.

The overall diagnostic accuracy of WCE within the group receiving angiography was 67 percent (12/18). The predictive value of a positive test was 100 percent (10/10). The predictive value of a negative WCE test was 50 percent, although this is based on just four negative WCE tests. The sensitivity of the WCE test was 63 percent (10/16). Only two patients referred for angiography were found to be free of significant CAD. Both of these individuals had negative WCE tests (specificity = 100 percent). However, because of the small number of cases, a meaningful evaluation of WCE test specificity is not possible.

Table 2.
Angiography evaluation of subjects by wheelchair ergometer (WCE) test results.

WCE Test Result	No Significant CAD	One Vessel CAD	Two Vessel CAD	Three Vessel CAD
Positive	0	5	3	2
Negative	2	1	1	0
Non-Diagnostic	0	0	3	1

WCE = wheelchair ergometer; CAD = coronary artery disease

DISCUSSION

In the present study, WCE exercise was utilized for diagnostic testing of persons (most with lower limb disabilities) who had known or suspected CAD. This is, to the authors' knowledge, the first research to evaluate WCE exercise for this purpose.

Upper Body Exercise Testing for the Diagnosis of Coronary Artery Disease

Most investigators have found that upper body exercise testing is less sensitive than treadmill or cycle ergometer testing for the detection of CAD (6,11,15,16,27). Generally, upper body exercise elicits lower peak cardiovascular stress, though a few researchers have reported equivalent responses (12,14,18). There appears to be a consensus that upper body exercise provides an acceptable, but less sensitive, alternative to lower body exercise for testing patients who cannot adequately perform these modes of exercise (6,11,12,14,16–18).

Comparison of Arm Crank Ergometry and Wheelchair Ergometry

Peak $\dot{V}O_2$ has consistently been found to be similar whether measured with WCE or arm crank ergometry (28). Some investigators have concluded that WCE exercise elicits a lower peak heart rate than arm crank ergometry (28). However, this has not been a universal finding (29,30) and is not supported by a previous study conducted in the authors' laboratory (21,22,31). In that investigation, maximal WCE and arm crank ergometry tests were completed by a sample of apparently healthy lower limb disabled men,

most with lower limb paralysis due to spinal cord injury. The sample was divided into three groups, upper-, mid-, and lower level injury. Within each group, mean peak rate-pressure product, heart rate, and $\dot{V}O_2$ were equivalent or greater for WCE than arm crank ergometry (21). Peak rate-pressure product was slightly but significantly greater for WCE (221 ± 71 vs. $214 \pm 67 \times 10^2$; $p = 0.05$) when the combined data from the entire group was analyzed. Peak $\dot{V}O_2$ also tended to be greater with WCE (1.34 ± 0.46 vs. 1.29 ± 0.43 L·min^{-1}; $p = 0.06$). Because the subjects were all manual wheelchair users, higher peak values for WCE may reflect specificity of training and/or greater task familiarity. These observations suggest that WCE is at least equivalent to arm crank ergometry for the detection of CAD, and may be preferable for manual wheelchair users.

Tables 3 and **4** summarize a number of previously published arm crank ergometry investigations. The mean peak heart rate (132 ± 26 b·min^{-1}, 84 ± 15 percent of APMHR) and rate-pressure product (223×10^2) achieved with WCE in the current study were not unlike those reported for arm crank ergometry in comparable patient samples. Several factors must be considered in contrasting results of upper body exercise studies, including patient profile (sex, age, presence of CAD or other conditions), use of medications which could influence results (e.g., beta adrenergic blockers) and method of blood pressure measurement (intra-arterial or auscultation).

Differences in peak upper body exercise responses between persons with CAD or peripheral vascular disease and healthy age-matched individuals are illustrated by two recent investigations. Manfre et al. (15) compared the arm crank ergometry and treadmill responses of 19 men with documented CAD (mean age 58 yr) and 12 healthy men (mean age 56 yr). Peak percent of APMHR (96 vs. 81 percent) and rate-pressure product (262×10^2 vs. 215×10^2) during arm crank ergometry were significantly greater for the healthy men. Goodman et al. (11) compared the upper body ergometry (Schwinn Air Dyne) responses of

Table 3.
Sample demographics and test protocols of arm crank exercise tests from selected published investigations, 1965 to present.

Reference	Year	Subject Profile	Mean Age (yr)	Age Range	Stage (min)	Rest period
Present study	1992	Known/suspected CAD (n = 49)	63 ± 9	42–81	3	none
Blomqvist[32]	1965	Angina; 5 known CAD (n = 6)	57	46–49	N/A	none
Wahren[33]	1971	Signs of CAD (n = 10)	52 ± 12	25–65	6	none
Shaw[18]	1974	Ambulatory (a) (n = 21)	57	41–74	3	none
		Non-ambulatory (b)(c) (n = 26)				
Schwade[17]	1977	MI or suspected CAD (n = 33)	52	42–67	3	1 min
DeBusk[9]	1978	Known MI (d) (n = 40)	51 ± 7	34–63	3	1 min
Lazarus[13]	1981	Angina (e) (n = 11)	58	48–64	3	none
Balady[6]	1985	Known CAD (f) (n = 30)	59 ± 9	40–71	3	none
Balady[7]	1987	Known/suspected CAD (n = 50)	56 ± 10	37–77	2	none
Fletcher[10]	1988	Musculoskeletal (g) (n = 15)	65 ± 13	39–84	3	1 min
Hanson[12]	1988	PVD, male (n = 57)	61 ± 10		2	2 min
		PVD, female (n = 17)	63 ± 10			
Goodman[11]	1989	PVD (n = 32)	62 ± 8		3	none
		Healthy (n = 17)	59 ± 10			
Balady[8]	1990	Healthy (n = 20)		40–59	2	none
Levandoski[14]	1990	History of CAD (n = 21)	60 ± 7		3	none
Manfre[15]	1990	Healthy (n = 12)	57		2	none
		Known CAD (g) (n = 19)	58			
Cullinane[27]	1992	Healthy (n = 9)		50–59	1	none
		Healthy (n = 7)		60–69		
Sala[16]	1992	Previous MI, male (n = 108)	48 ± 7		3	none
		Previous MI, female (n = 20)	50 ± 6			
Davidoff[20]	1992	Dysvascular Amputees (n = 25)	63 ± 8		2.5	0.5 min

(a) all MI or abnormal angiogram; (b) 14 peripheral vascular disease, 4 arthritis, 2 amputees, 4 hip prostheses, 2 neurological disease; 7 MI or abnormal angiogram; (c) no digitalis within 10 days; no propranolol within 48 hours; (d) no subjects using digitalis; all medications discontinued on day of test; (e) no subjects using beta blockers; (f) no subjects using digitalis; (g) beta blockers discontinued 12–24 hours before testing. MI = myocaridal infarction; CAD = coronary artery disease; PVD = peripheral vascular disease.

Table 4.
Comparison of peak cardiovascular responses to arm crank exercise tests from selected published investigations, 1965 to present.

Reference	Systolic BP (mmHg)	Heart Rate ($b \cdot min^{-1}$)	Percent APMHR	RPP ($\times 10^2$)	$\dot{V}O_2$ ($L \cdot min^{-1}$) [$mL \cdot kg^{-1} \cdot min^{-1}$]
Present study	169 ± 29	132 ± 26	84.1 ± 14.6	223 ± 62	1.13 ± 0.4; n = 23 [14.20 ± 4.5; n = 23]
Blomqvist[32]	218 ± 34 (a,b)	120 ± 20 (b)			0.87 ± 0.4
Wahren[33]	207 ± 23 (a)	134 ± 23	79.9 ± 11.6	277 ± 57	
Shaw[18]	157 ± 7		81.0 ± 4.0	220 ± 12	
	167 ± 8		73.2 ± 1.9	224 ± 12	
Schwade[17]		122		234	
DeBusk[9]	158 ± 12	150 ± 11		236 ± 21	
Lazarus[13]	194 ± 19	127 ± 17		246 ± 43	0.99 ± 0.1
Balady[6]	161 ± 20	101 ± 18		160 ± 40	[13.00 ± 5.0]
Balady[7]		114 ± 22	70.0 ± 14.0	180 ± 48	
Fletcher[10]	127 ± 28	106 ± 22	(c)		
Hanson[12]	167 ± 34	142 ± 22	91.0 ± 14.0	239 ± 62	
	161 ± 38	132 ± 22	86.0 ± 13.0	211 ± 52	
Goodman[11]	197 ± 31	138 ± 20	77.1 ± 10.7	274 ± 68	[17.90 ± 4.4; n = 25]
	176 ± 20	153 ± 16	85.6 ± 8.9	271 ± 49	[22.00 ± 4.1; n = 15]
Balady[8]			91.0 ± 10.0		[18.30 ± 4.5]
Levandoski[14]		144 ± 23			1.63 ± 0.4
Manfre[15]	167		95.7	262	[21.40]
	165		80.7	215	[15.00]
Cullinane[27]	160 ± 20	149 ± 15		239 ± 43	1.63 ± 0.2
	161 ± 31	136 ± 12		222 ± 58	1.27 ± 0.3
Sala[16]	164 ± 24	130 ± 21	(d)	212 ± 45	
	171 ± 20	126 ± 17		215 ± 36	
Davidoff[20]	152 ± 29	119 ± 13		182	

(a) intra-arterial measurement; (b) measurement at onset of pain; (c) exercise target 75% of maximum; (d) exercise target 85% of maximum. BP = blood pressure; APMHR = age predicted maximal heart rate; RPP = rate pressure product; $\dot{V}O_2$ = oxygen uptake.

32 men with peripheral vascular disease (61.6 ± 7.8 yr) to 17 healthy age-matched controls. Peak percent of APMHR was lower for the peripheral vascular disease patients (77 vs. 86 percent). Because of a higher peak systolic blood pressure in the peripheral vascular disease patients (197 vs. 176 mmHg), peak rate-pressure product was similar (274 $\times 10^2$ vs. 271 $\times 10^2$). In comparison, Hanson et al. (12) obtained a mean peak heart rate of 142 $b \cdot min^{-1}$ (91 percent APMHR) and a mean peak rate-pressure product of 239 $\times 10^2$ in 57 men with peripheral vascular disease (61 ± 10 yr), whereas Davidoff et al. (20) reported a mean peak heart rate of 119 $b \cdot min^{-1}$ and rate-pressure product of 182 $\times 10^2$ in 25 dysvascular amputees (63 ± 8 yr).

Balady and colleagues (6,7) published two investigations in which arm crank ergometry was used to detect myocardial ischemia in patients with angina pectoris. Nearly all subjects in both studies were taking antianginal medications. Relatively low peak exercise responses were obtained, possibly secondary to the use of antianginal medications and/or the presence of signs and symptoms of CAD which limited exercise performance. Peak heart rate and rate-pressure product in the two reports were 101 and 114 $b \cdot min^{-1}$ and 160 $\times 10^2$ and 180 $\times 10^2$, respectively. Sala et al. (16) reported on arm crank ergometry tests performed in a large group of post-myocardial infarction patients, 108 men (48 ± 7 yr) and 20 women (50 ± 6 yr). As compared to the results of the present investigation, the mean peak heart rate (130 $b \cdot min^{-1}$ for men, 126 $b \cdot min^{-1}$ for women) and rate-pressure product (212 $\times 10^2$ for men, 215 $\times 10^2$ for women) were very similar.

Blood pressures measured by auscultation that are reported in this study, and by others, probably underestimate the true systolic pressure (and therefore rate-pressure product) during upper body exercise. Systolic blood pressures measured intra-arterially during arm crank ergometry (32,33) were substantially greater than those reported in studies utilizing the standard cuff method (**Tables 3** and **4**). Hollingsworth et al. (34) compared systolic blood pres-

sures taken by auscultation immediately upon cessation of each arm crank exercise stage with those estimated by using a Doppler flowmeter technique during exercise. They found that measurements taken immediately post-exercise underestimated the exercise pressure by 7–22 percent. The difference was greatest at higher workloads.

Mean peak $\dot{V}O_2$ for the 23 patients in the current study who completed metabolic testing was 1.13 ± 0.4 L·min^{-1} (14.2 ± 4.5 mL·kg·min^{-1}). This is higher than that observed by Blomqvist et al. (32) (0.87 L·min^{-1}, n = 6) and Lazarus et al. (13) (0.99 L·min^{-1}, n = 11) for arm crank ergometry by patients with angina or known CAD. Balady et al. (6) and Manfre et al. (15) reported the arm crank peak $\dot{V}O_2$ of patients with CAD to be 13.0 mL·kg·min^{-1} (n = 30) and 15.0 mL·kg·min^{-1} (n = 19), respectively. These values are similar to the peak aerobic capacity during WCE in the present investigation. Greater average peak $\dot{V}O_2$ measures were found by Levandoski et al. (14) (1.63 L·min^{-1}, n = 21) in patients with CAD performing arm crank ergometry and by Goodman et al. (17.9 mL·kg·min^{-1}, n = 25) in peripheral vascular disease patients during upper body ergometry on a Schwinn Air Dyne (11).

Upper body exercise tests are frequently terminated before significant cardiovascular stress is achieved due to local fatigue resulting from the use of a relatively small working muscle mass. This is considered a major limitation to the use of these modes of exercise for detection of CAD (19). By conventional criteria (35), most subjects in the present study were able to reach a clinically acceptable level of cardiovascular stress for diagnostic purposes, despite the fact that many were taking medications which would tend to limit the chronotropic response (beta receptor and calcium channel blockers). Sixty-nine percent of patients attained at least 85 percent of APMHR and/or had positive WCE tests. To account for the lower peak heart rate expected with upper body work, target heart rate for a diagnostic test was defined as 80 percent of APMHR (26,29). With this target, 76 percent of the subjects were able to acheive the target heart rate or had positive test results.

Coronary Angiography

The predictive value of a positive WCE test was 100 percent (10/10) for the subset of 18 subjects for whom angiographic data were obtained. This value is consistent with arm crank ergometry results reported by Travers et al. (19) (100 percent), Hanson et al. (12) (88 percent overall, 91 percent for men only), and Balady et al. (6) (100 percent). The predictive value of a negative WCE test was 50 percent, although this finding was based on only four angiograms. Travers et al. (19) found that the predictive value of a negative arm crank ergometry test was 37 percent. Again, this was based on a small number of angiograms (n = 9). The sensitivity and diagnostic accuracy of WCE testing for the patients who underwent angiography were 63 percent and 67 percent, respectively. These values were undoubtedly influenced by the small number of angiograms and selection bias in referral for angiography in that 72 percent of patients who underwent angiography had positive WCE tests. Test specificity could not be meaningfully evaluated because of the small number of subjects without significant CAD who underwent angiographic assessment (n = 2). These subjects both had negative WCE tests (specificity = 100 percent).

The most commonly employed noninvasive methods of testing for CAD in the lower limb disabled utilize thallium or echocardiographic imaging with pharmacologic stress (e.g., adenosine or dobutamine). These methods have been demonstrated to possess high degrees of sensitivity and specificity (36,37). Notable disadvantages associated with pharmacologic testing include drug side effects, exposure to radiation if radionuclide imaging is used, and expense. Also, pharmacologic stress provides no information regarding a patient's functional capacity. Based on the high predictive value of a positive WCE or arm crank ergometry test and the relative disadvantages associated with pharmacologic tests, it would be reasonable to first utilize WCE or arm crank ergometry testing in some patients for whom lower body exercise would not be an option. This might be a particularly useful and cost-effective strategy for those with a high pretest likelihood of disease.

Effects of Cardiovascular Medications on WCE Results

Antianginal medications were taken by 63 percent of the subjects at the time of WCE testing, which may have reduced sensitivity for the detection of ischemia. No significant differences were found in peak rate-pressure product, heart rate, or systolic blood pressure between patients taking antianginals and those patients not on these medications. Similarly, no differences were detected in these parameters between patients taking beta blockers and the remainder of the sample. The distribution of WCE test results was also independent of whether or not subjects were taking antianginals or beta-blockers ($p > 0.05$). Nevertheless, it is possible that a greater proportion of tests

would have been positive had these medications been discontinued before testing.

Four of the ten patients taking digitalis had significant ST segment depression. These tests were considered non-diagnostic because digitalis is known to increase the rate of false positive ST segment depression (25). Angiography was performed on three of these subjects and confirmed the presence of significant CAD in all of them (two with double- and one with triple vessel-disease). Had these tests been considered positive, the diagnostic accuracy and sensitivity of WCE would have been 83 percent and 81 percent, respectively.

Study Limitations

Caution is recommended in generalizing the findings of the present study to other patient groups. Angiographic data could only be obtained on a limited subset of subjects and there was a probable selection bias in referral for WCE testing and for angiography. Many patients also had a high pretest likelihood of disease based on their age, sex, and the presence of concurrent conditions (e.g., peripheral vascular disease, lower limb amputations) (3–5,25,38). However, the authors believe that this group was a fairly representative cross section of the veteran population with lower limb disablility and known or suspected CAD.

For Further Investigation

The current findings, in agreement with studies utilizing arm crank ergometry (6,12,19), provide evidence that a positive upper body exercise test is highly predictive of significant CAD. In order to assess the cost effectiveness of WCE as a diagnostic tool, further study will be required to determine the sensitivity, specificity, and the predictive value of a negative test. This will require direct comparison of WCE results with those obtained by coronary angiography in a larger sample of patients. The prognostic significance of upper body exercise capacity has also not been addressed. WCE should be compared directly to arm crank ergometry for detecting CAD in wheelchair users and non-users to assess the possible advantages related to specificity of training and task familiarity. In addition, research to evaluate the combined effectiveness of thallium imaging or echocardiography in conjunction with WCE exercise testing is indicated. Balady et al. (7) compared arm crank ergometry with and without thallium scintigraphy. Detection of significant CAD by ECG criteria alone showed a sensitivity of 54 percent and specificity of 67 percent. Thallium imaging improved the sensitivity to 83 percent and specificity to 78 percent. This suggests that, in some patients, ischemia may occur during upper body exercise that is not severe enough to be detected by ECG alone. Therefore, thallium scintigraphy or echocardiography may improve the sensitivity and specificity of upper body exercise testing. The observations of Goodman et al. (11), Manfre et al. (15) and Bauman et al. (39) also support this possibility.

CONCLUSIONS

The findings of this initial evaluation provide evidence that WCE exercise testing is at least equivalent to arm crank ergometry for detection of CAD. Peak rate-pressure product and percent of APMHR were similar to those reported for arm crank ergometry in comparable groups of patients, and a majority of subjects were able to achieve a clinically acceptable level of stress for diagnostic purposes. Furthermore, the high predictive value of a positive test suggests that WCE testing is a viable initial diagnostic alternative to pharmacologic stress for some patients (e.g., those with a high pretest likelihood of disease). As compared to pharmacologic stress, WCE is less costly, involves no exposure to radiation or intravascular access, has fewer side effects, and provides valuable information about patients' functional capacity, particularly for wheelchair users. Further research will be required to establish WCE test sensitivity, specificity, as well as the predictive value of a negative test and its usefulness as a prognostic tool.

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REFERENCES

1. Brenes G, Dearwater S, Shapera R. High density lipoprotein cholesterol concentrations in physically active and sedentary spinal cord injured patients. Arch Phys Med Rehabil 1986:67:445–50.
2. Gersh B, Charanjit R, Rook T, Ballard D. Evaluation and management of patients with both peripheral vascular and coronary artery disease. J Am Coll Cardiol 1991:18:203–14.

3. Hrubec Z, Ryder R. Traumatic limb amputations and subsequent mortality from cardiovascular disease and other causes. J Chron Dis 1980:33:239–50.
4. Smith D, Shipley M, Rose D. Intermittent claudication, heart disease risk factors, and mortality: The Whitehall study. Circulation 1990:82:1925–31.
5. Yekutiel M, Brooks M, Ohry A, Yarom J. The prevalence of hypertension, ischaemic heart disease and diabetes in traumatic spinal cord injured patients and amputees. Paraplegia 1989:27:58–62.
6. Balady G, Weiner D, McCabe C, Ryan T. Value of arm exercise testing in detection of coronary artery disease. Am J Cardiol 1985:55:37–9.
7. Balady G, Weiner D, Rothendler J, Ryan T. Arm exercise-thallium imaging testing for the detection of coronary artery disease. J Am Coll Cardiol 1987:9:84–8.
8. Balady G, Weaver D, Rose L, Ryan T. Physiologic responses to arm ergometry exercise relative to age and gender. J Am Col Cardiol 1990:130–5.
9. DeBusk R, Valdez R. Cardiovascular responses to dynamic and static effort soon after myocardial infarction. Circulation 1978:58:368–75.
10. Fletcher G, Lloyd A, Waling J, Fletcher B. Exercise testing in patients with musculoskeletal handicaps. Arch Phys Med Rehabil 1988:69:123–7.
11. Goodman S, Rubler S, Bryk H, Sklar B, Glasser L. Arm exercise testing with myocardial scintigraphy in asymptomatic patients with peripheral vascular disease. Chest 1989:95:740–6.
12. Hanson P, Pease M, Berkoff H, Turnipseed W, Detmer D. Arm exercise testing for coronary artery disease in patients with peripheral vascular disease. Clin Cardiol 1988:11:70–4.
13. Lazarus B, Cullinane E, Thompson P. Comparison of the results and reproducibility of arm and leg exercise tests in men with angina pectoris. Am J Cardiol 1981:47:1075–9.
14. Levandoski S, Sheldahl L, Wilke N, Tristani F, Hoffman M. Cardiorespiratory responses of coronary artery disease patients to arm and leg cycle ergometry. J Cardiopul Rehabil 1990:10:39–44.
15. Manfre M, Mallis G, Varma A, Yen O, Plon M. Oxygen uptake, electrocardiographic, and symptomatic response during two arm cranking and treadmill testing in healthy older adult men and patients with coronary artery disease. J Cardiopul Rehabil 1990:10:12–20.
16. Sala R, Laitinen L. Comparison of arm crank exercise with leg ergometry in the evaluation of patients with past myocardial infarction. J Cardiopul Rehabil 1992:12:174–82.
17. Schwade J, Bloomqvist C, Shapiro W. A comparison of the response to arm and leg work in patients with ischemic heart disease. Am Heart J 1977:94:203–8.
18. Shaw D, Crawford M, Karliner J, et al. Arm-crank ergometry: a new method for the evaluation of coronary artery disease. Am J Cardiol 1974:33:801–5.
19. Travers A, Nel C, Van Der Watt F, Jordaan P. The arm ergometer exercise test for evaluating coronary artery status in patients presenting for peripheral vascular surgery. S Afr J Surg 1990:28:148–50.
20. Davidoff G, Lampman R, Westbury L, et al. Exercise testing and training of persons with dysvascular amputation: safety and efficacy of arm ergometry. Arch Phys Med Rehabil 1992:73:334–8.
21. Langbein WE, Fehr L. Research device to preproduction prototype: a chronology. J Rehabil Res Dev 1993:30(4):436–42.
22. Langbein WE, Robinson CJ, Kynast L, Fehr L. Calibration of a new wheelchair ergometer: the wheelchair aerobic fitness trainer. IEEE Trans Rehabil Eng 1993:1:49–58.
23. The Preventive and Rehabilitative Exercise Committee of the American College of Sports Medicine. Guidelines for graded exercise testing and exercise prescription. 4th ed. Philadelphia: Lea & Febiger, 1986.
24. Fletcher G, Froelicher V, Hartley L, et al. Exercise standards: a statement for health professionals from the American Heart Association. Circulation 1990:82:2286–322.
25. Ellestad M. Stress Testing. 3rd ed. Philadelphia: F.A. Davis Co., 1986.
26. Franklin B. Exercise testing, training and arm ergometry. Sports Med 1985:2:100–19.
27. Cullinane E, Ribeiro B, Sady S, Miller F, Thompson P. Can treadmill exercise capacity be predicted from arm ergometry results? J Cardiopul Rehabil 1992:12:36–41.
28. Glaser R, Davis G. Wheelchair-dependent individuals. In: Franklin BA, Gordon S, Timmis GC, eds. Exercise in modern medicine. Baltimore: Williams & Wilkins, 1989:237–67.
29. Sawka M. Physiology of upper body exercise. Exer Sport Sci Rev 1986:14:175–211.
30. Shephard R. Fitness in special populations. Champaign, IL: Human Kinetics Books, 1990.
31. Langbein W, Robinson C, Hwang M, et al. Validation of a new wheelchair ergometer for graded exercise testing of persons with lower limb disabilities. Circulation 1989:80(suppl II):II–247.
32. Blomqvist G, Astrand I, Messin R. Clinical, physiological, and electrocardiographic findings in patients with angina pectoris during work in cold environment and during arm work. Acta Med Scand 1965:178(suppl 440):74–81.
33. Wahren J, Stellan B. Onset of angina pectoris in relation to circulatory adaptation during arm and leg exercise. Circulation 1971:44:432–41.
34. Hollingsworth V, Bendick P, Franklin B, et al. Validity of arm ergometer blood pressures immediately after exercise. Am J Cardiol 1990:65:1358–60.
35. Franklin B, Hellerstein H, Gordon S, Timmis G. Cardiac patients. In: Franklin BA, Gordon S, Timmis GC, eds. Exercise in modern medicine. Baltimore: Williams & Wilkins, 1989:44–80.
36. Beller G. Current status of nuclear cardiology techniques. Curr Probl Cardiol 1991:451–535.
37. Picano E. Stress echocardiography: from pathophysiological toy to diagnostic tool. Circulation 1992:85:1604–12.
38. Weiner D, Ryan T, McCabe C, Kennedy J, Schloss M, Tristani F, et al. Exercise stress testing: correlations among history of angina, ST segment response and prevalence of coronary artery disease in the coronary artery surgery study (CASS). N Engl J Med 1979:301:230–5.
39. Bauman WA, Spungen AM, Raza M, et al. Coronary artery disease: metabolic risk factors and latent disease in individuals with parpaplegia. Mt Sinai J Med 1992:59:163–8.

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The Southampton Hand: An intelligent myoelectric prosthesis

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Abstract—The form of the control and structure of the mechanism of an artificial hand are important factors which tend to dictate the prosthesis' level of use. Conventional prostheses are simple devices with limited functional range and a control format that requires high levels of user concentration for successful operation. The Southampton Adaptive Manipulation Scheme (SAMS) is a hierarchical control format that allows a larger number of independent motions to be controlled while requiring a smaller degree of user input. The SAMS control has been applied to different hand mechanisms, both custom-made and modified commercial systems. Their application with users shows them to have a performance on a par with, or superior to, other conventional devices. The form of prostehsis control is reviewed and the development of, and clinical experiments with, the Southampton Hand are outlined.

Key words: *artificial hand, microprocessor control, myoelectric prosthesis control, prehension.*

INTRODUCTION

A prostheses can be either actuated by an operator, using his/her own body to power the device, or it can derive its power from an external source. The control of the device is then dictated by this choice (1,2). Body-powered prostheses use the direct mechanical link between the operator and the device to control them. Good control is possible, and complex manipulation can be achieved if an effective design is employed. Generally, the commercial designs that are functional are not aesthetic, although this is now beginning to change (3,4), and functional hands are never anthropomorphic. In addition, actuation is generally bulky, requiring straps or cables which chafe against the clothes and become dirty. The range of movement or power of the device can be compromised by the geometry of the system. Despite these drawbacks, bodily powered prostheses are the most common form of terminal device, due to their functional range, robustness, light weight, and low cost.

The only practical external power source is electric; this is due to the ease by which the power source can be recharged compared with the difficulties of recharging any other safe source (5,6). Electronics also provide a compact controller. The resulting device can be more cosmetic in appearance, needing no straps to open it and much smaller bodily actions to operate it; in addition, it is less tiring to use. However, the major feedback path from the device to the operator that exists with body powered prostheses is severed. It must either be reestablished or circumvented to allow good control of the device. Current electric hands do not use any feedback except visual and incidental forms of motor vibration (7), thus the control burden is higher in the electric hands than in the mechanical ones.

Some forms of feedback have been investigated, such as the coding of the force or hand flexion in vibrations (8,9) but this mental transformation is burdensome. If the feedback is applied to the correct point in an appropriate manner by using, for example, extended physiological proprioception (EPP), good tracking performance of an arm can result (10–13). The shortcoming of such a method is that it works well for the large actions that control a serial line of joints (for example, in an arm), but cannot as easily control the parallel joints in a hand mechanism.

Commercial artificial hands have a single degree of freedom that allows them to open and close. The cosmetic versions generally are anthropomorphically shaped hands in the form of a precision grip. They cannot open wide enough to admit many common objects or flex far enough

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to hold small objects in a power grip. This limits their functional range. Although different manufacturers' hands have subtly different geometries, they do not vary greatly in prehensile performance. All commercial hand-like devices have the same limitations on size and grip forms. The standard configuration is an anthropomorphic hand in a three-point chuck grip (14,15). These can hold large objects in a power grip or precision-type grip, but do not perform so well with the smaller objects (5,16). Changes in hand geometry to a more anthropomorphic form circumvent this drawback as the fingers can curl around small objects, or alternatively, the tips of the index finger and thumb can oppose each other (17–19). Even more adaptive is the hand where the thumb can abduct to oppose the side of the index finger (20,21). However, the matter of the control of the device still is problematical.

In principle (everything else being equal), increased function could come from an increased number of independent motions, but conventional control requires too much concentration from the user, needing one independent input channel for each degree of freedom. An alternative method, developed in the Department of Electrical Engineering at the University of Southampton, mimics the natural control method of the human hand and so frees the user from making detailed decisions about the hand (22,23).

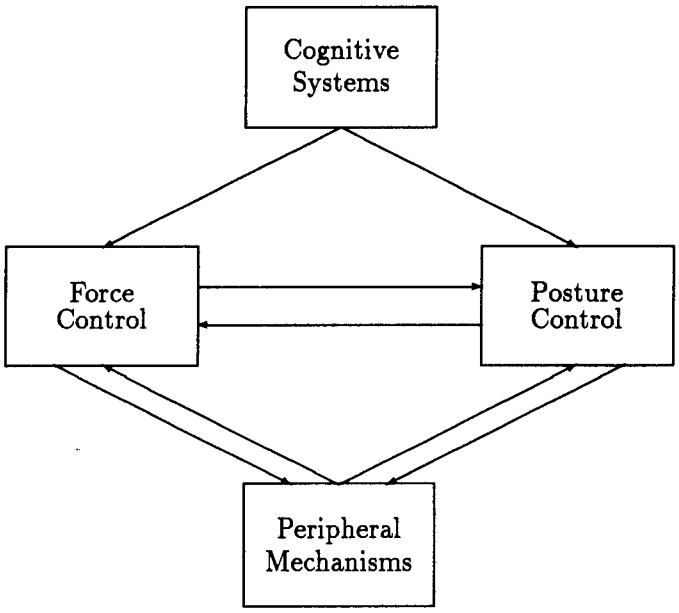

Figure 1.
The Southampton Adaptive Manipulation Scheme, (SAMS). It is arranged in a hierarchical form analogous to the human Central Nervous System.

METHOD

The Southampton Adaptive Manipulation Scheme (SAMS) was developed to address the problems related to the control of the multi-degree-of-freedom and multifunction hand prosthesis. The basic form of the central nervous system (CNS) of a human being is hierarchical, as the tasks of controlling the hand and digits are broken up into three layers (**Figure 1**). At the lowest level the force and position of individual fingers are managed. These reflexes are then commanded by an intermediate level that coordinates the fingers to create a hand shape and grip force in response to the shape of the target object and the action that is intended for it. Above this is the strategic control of the hand. This is the level of the consciousness of the individual. The person simply desires to move an object, and the system coordinates the action to achieve this goal with very little conscious thought. SAMS was designed to restore the level of control of a prosthesis up to this level (**Figure 2**).

The user issues simple instructions to open the hand, normally through a single electromyographic (EMG) channel. In this example, the flexor and extensor muscles on the forearm are used (**Figure 3**). The EMG channel is regarded as a single bipolar signal ranging from full extension at one extreme to full flexor tension at the other. In the center both muscles are relaxed. The degree of opening is proportional to the muscular tension (on the positive vertical axis); therefore, when the muscle relaxes, the hand closes natu-

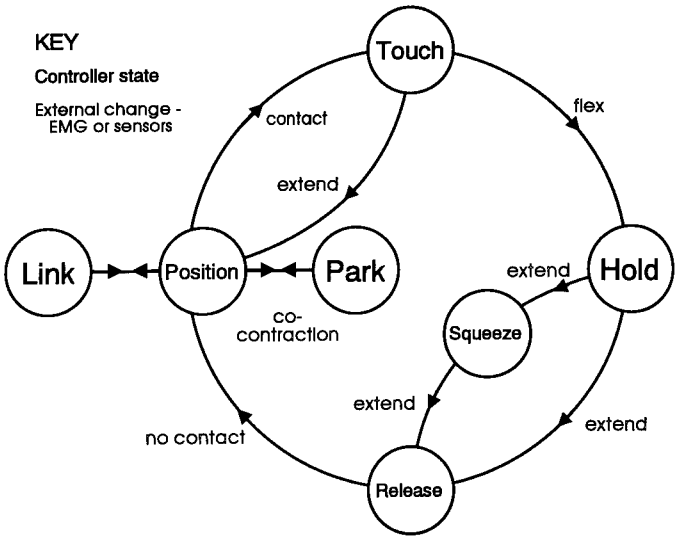

Figure 2.
State diagram of the SAMS hands. Control is mediated by electromyographic input or contact with sensors on the palmar surface of the hand.

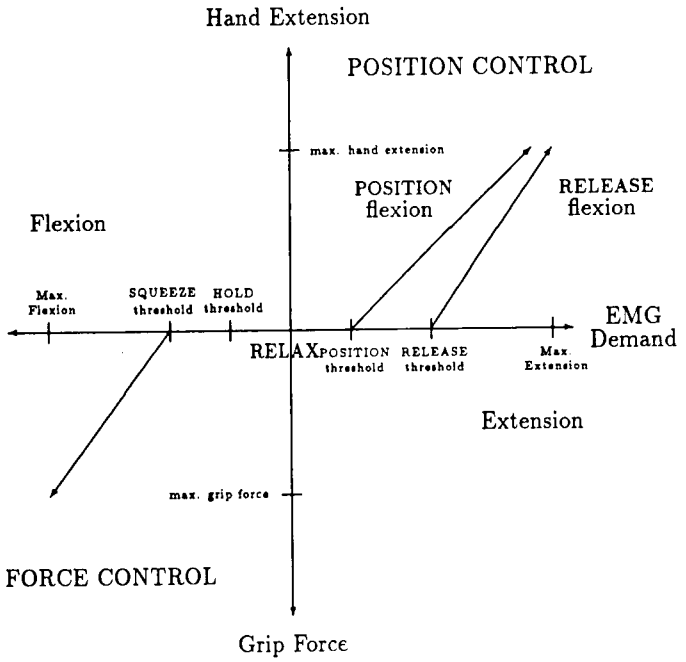

Figure 3.
EMG control for the Southampton Adaptive Manipulation Scheme (SAMS). The flexor and extensor signals are arranged as a continuous range from full flexion to full extension, the vertical axis is hand position or grip for demand, depending on controller state.

rally upon the object (POSITION). The shape of the object is detected by sensors on the palmar surface of the hand while a computer controller selects a grip posture from a small repertoire to suit the most appropriate general shape. The controller then makes detailed corrections of that shape to suit the exact shape of the object. This maximizes the contact area while minimizing the contact force. In this phase, a light touch is maintained so the operator can maneuver the object within the hand to obtain the best attitude (TOUCH). Then the user can instruct the computer to hold the object (HOLD). If the grip tension is too low, the object slips within the grasp, the slippage is detected by sensors on the hand, resulting in an increase in the force in proportion to the time that slippage occurs. At any time, the operator can either instruct the hand to increase the grip force, overriding the slip reflex (SQUEEZE, the negative going arm of the y-axis, **Figure 3**) or to open (RELEASE). The threshold when this occurs can be set higher than for when the hand is opened empty, so that holding or releasing objects becomes a more deliberate act than opening the hand when empty. With conventional prostheses this point has to be set at the same muscular tension so that it is either too easy to drop an object, or too difficult to use the hand without tiring. A LINK phase can also be implemented in the SAMS scheme so that the third, fourth, and fifth fingers are moved out of the way and the hand adopts a two-digit pinch grip for manipulating small objects. Finally, a PARK state allows the hand to be powered down when not in use.

This scheme has been realized on a range of prostheses known generically as the "Southampton Hand." The Southampton Hand is the entire system, comprising the adaptive control scheme, the proportional instructions controlling a terminal device with feedback to the controller. Analysis of the type of action required to perform the majority of prehension tasks showed that for an anthropomorphic hand, four degrees of freedom are sufficient (20). These motions are: index finger flexion and extension; thumb flexion/extension and abduction/adduction; and flexion of the other three digits as a closely coupled group. Using the terminology described by Napier (24), the design allows the hand to perform *precision* prehension with two or more digits, *power* grip, as well as *side* prehension, where the thumb opposes the side of the index finger. The hand was designed to look and move in an anthropomorphic way to provide a pleasing cosmesis for the device. It was also designed to fit all four drives within the natural envelope of the hand so it could be worn by the widest range of possible users.

The slip detection and response realized on the Southampton Hand is based on detecting the vibrations set up when an object slides past the finger tips (23), or the changes in the forces between the hand and the object (7). These techniques mirror those of natural systems (25,26). The advantages of the vibrotactile detection have been recognized and are being implemented in a number of other devices (18,19,27,28).

Early Clinical Demonstration

The original Southampton Hand was built in 1969 (20) and was controlled by using discrete logic components. A realistic clinical application requires the electronics to be reduced to a size and cost that is acceptable to the user population and the limb-providing authorities. In the early 1970s, the electronic devices and packaging technology were not sufficiently small to allow the production of a clinically practical device. In the interim, laboratory versions of the hand were fitted to an individual who usually used a split hook (Dorrence heavy duty hook, number 7, voluntary opening configuration). He had a congenital, below-elbow right hand loss (29). A conventional myoelectric hand (Viennatone MM3 with digital

control of velocity and force) was fitted, so a comparison of the control methodologies could be made (**Figure 4**).

The subject was trained prior to performing various tests to assess his ability to use the hand. The provision of the myoelectric hand also allowed the subject to accustom himself to myoelectric control between training sessions and prior to the laboratory tests. The procedure was in the form of progressively introducing more of the functions to him, spread over a number of visits to the center:

1. Fitting of prosthesis.
2. Familiarization of EMG scheme without hand.
3. Familiarization of hand functions.
4. Grasping of everyday objects.
5. Evaluation:

- Abstract prehension objects
- Positional prehension on standard objects on open shelves
- Practical activities outlined in **Table 1**.

Familiarization of EMG Scheme

Initially, an oscilloscope trace of the smoothed EMG output was displayed for the subject and used to indicate the command level achieved. Once the subject was sure of the operation, this was replaced by lights that indicated the state the controller was in. This was found to be a very clear way to train the user. Once the error rate was below 6 percent (after 2.5 hours), the subject progressed to the next phase.

Familiarization of Hand Functions

Familiarization was achieved by separating the functions into each grip category and progressively introducing them by use of a range of abstract objects. Practical tasks are difficult to assess objectively; therefore, the abstract objects were adopted. Compressible foam items were designed to test the force control. These were made of strips of plastic foam glued together at their centers only. The foam was a light color with the ends colored a darker shade. Thus, pressure on the central region caused the strips to splay out progressively relative to the force imparted, betraying the level of force to the outside observer. Competence was achieved at these tasks after a further 4.25 hours.

Evaluation

Prior to the start of the evaluation, the subject was advised that he would be scored on the time taken to complete the task and the number of grasp errors. No period of practice was allowed before the first run of the test. This was to enable a measure of the competence of the control to be made. Any subsequent improvement in the times would betray this fact. The positional tasks consisted of moving abstract objects, such as cylinders and blocks, from lower shelves to higher ones, or vice versa. The practical tests were based on ones devised by the Department of Health and Social Security (DHSS) in the United Kingdom to assess artificial limbs. They consisted of abstract prehension tasks and simulated real tasks of daily living. The tasks were recorded, observed by an independent, experienced observer, and timed (29); the results are outlined in **Table 1**. The following ratings were given to the hands based on the scores given by the observer:

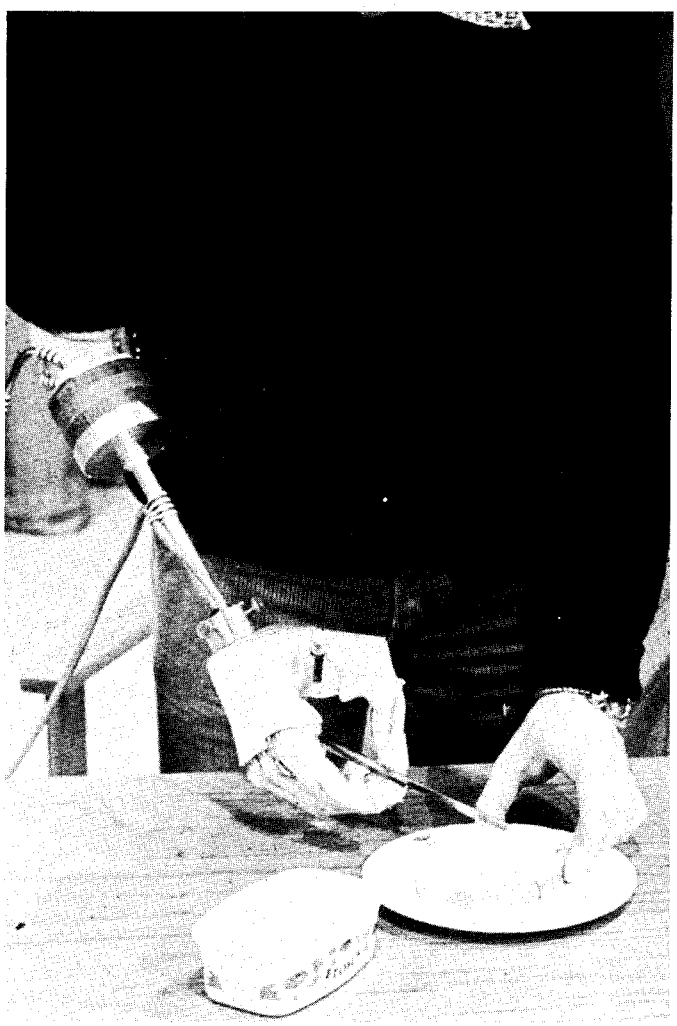

Figure 4.
Third generation four degree of freedom SAMS hand under test.

Table 1:
Comparative tests with the SAMS hand.

Task	Time(s)			Rating	
	Sams[1]	Hook[2]	Myo[3]	SAMS	Myo
Cutting					
Fork LH Knife RH	57	26	—	2	1
Fork RH Knife LH	49	42	—	2	1
Change grip, Spear to scoop	12	14	—	2	1
Open bottle and pour					
Top LH Bottle RH	26	29	—	3	1
Top RH Bottle LH	11	12	12	3	1
Carry tray	21	17	18	2	2
Cut slice of bread					
Loaf LH Knife RH	41	42	—	3	1
Loaf RH Knife LH	17	26	17	3	2
Butter bread					
Bread RH Knife LH	16	19	20	2	1
Bread LH Knife RH	36	31	29	3	2
Fasten belt	32	29	31	3	2
Toothpaste onto brush					
Brush LH Tube RH	36	21	20	3	2
Brush RH Tube LH	42	—	15	3	2
Grasp telephone receiver	19	5	5	3	2
Grasp pen and write	30	20	22	2	2
Cigarette from pack					
Pack LH Cig RH	28	20	44	2	2
Pack RH Cig LH	12	11	13	2	2
Use mallett and chisel					
Mallt LH Chis RH	16	11	9	3	3
Mallt RH Chis LH	18	—	15	3	3
Pick up coins	34	17	27	3	2
Lift and pour kettle	29	15	—	3	1
Tear and fold paper	46	46	26	2	2
Put paper in envelope					
Paper LH Env RH	19	18	22	2	2
Paper RH Env LH	13	18	20	2	2
Grasp cup	8	6	7	2	2

[1]A hierarchically controlled four degree of freedom Southampton prosthesis.
[2]Dorrence heavy duty split hook (no. 7), voluntary closing.
[3]Viennatone single degree of freedom hand with two myoelectric channels.

1. The hand was inferior to the split hook.
2. The hand was as successful as the split hook.
3. The hand was superior to the split hook.

The total assessment was spread over 3 months with a total of 31.25 hours use of the hand.

Recent Progress

More recent work has concentrated on two areas: The first is sensor design and signal processing of the input signal for improved performance and speed (30,31). The second is development of a simple Southampton Hand that could be used clinically (7,32).

The device was based on a Viennatone MM3 hand. The standard control electronics were removed and sensors added to detect object slip, contact force, and hand flexion angle. A simple microprocessor-based computer was built from CMOS low-power components and the device was fitted to an individual who had suffered a traumatic amputation of the left wrist and usually used a myoelectric hand (Steeper 2-channel myoelectric hand).

After three familiarization/training sessions at the Arm Training School at Queen Mary's University Hospital, Roehampton, London, UK, the individual was able to use the hand at home and at work. In addition, pick-and-place tests were performed on a standard set of shapes and the standard bimanual tasks used at the center. These were recorded on video tape and observed.

RESULTS

Four Degree of Freedom Hand

For trials on the multifunction hand, the subject was able to learn the operation of the hand quickly (within half an hour). A few hours of use then refined the familiarity with the controls still further. For the larger abstract prehension tests, all three devices worked equally well. As the items became smaller, the grasp limitations of the standard myoelectric hand were more pronounced. On the real objects, the myoelectric hand was not as easily or quickly used, as the wider range of shapes were more readily adopted by the Southampton Hand. In addition, wrist pro-supernation was not necessary as the adaptive shape of the grasp allowed these to be accommodated as well.

The split hook was limited to the largest size of its grasp and it was also very tiring for the user to repeatedly open the hook to a sufficiently wide gape to admit many of the large objects.

Neither of these drawbacks was noted with the Southampton Hand. The observers' assessment of the hand was that it was superior in performance to the hook in just under half of the practical activities (12 out of 25, or 48 percent) and it was equal in the rest. The standard myoelectric hand was seen to be on a par with, or worse than, the hook. The area where the Southampton Hand showed the greatest advantage over the split hook was where the actions required a power grip with a large grasp. However, some of the other tasks were not possible at all (**Table 1**). The user was allowed to practice with the training objects until his times between runs tended toward a constant value. However, there were improvements in the times for the practical tasks, showing further improvement was still taking place.

Single Degree of Freedom SAMS Hand

These results were borne in mind when the experiment with the single degree of freedom Southampton Hand was commenced. Since the functional range of the device was obviously limited, the comparisons were made in terms of the ease of teaching and use of the device. The user's own device was a Steeper myoelectric hand with two channel digital input. Given that the hands were very similar in design and construction, the differences in use could be more directly attributable to the control philosophy than in the previous experiment.

The individual found the Southampton scheme easy to learn. However, the difference in control between the Southampton Hand and his normal prosthesis needed to be explained. Habit had taught him to use flexor tension to close the hand and extensor tension to open it. The Southampton Hand is of the normally closed form with the opening on the extensor tension alone; the flexor is used for switching to the hold state and SQUEEZE override, once HOLD is established. Only a little practice was required before he began to allow the hand to close of its own accord rather than to instruct it like his usual hand. Once this was achieved, he easily used the flexor tension to invoke HOLD or SQUEEZE. The subject became an enthusiastic user of the device and readily appreciated its advantages (**Figure 5**).

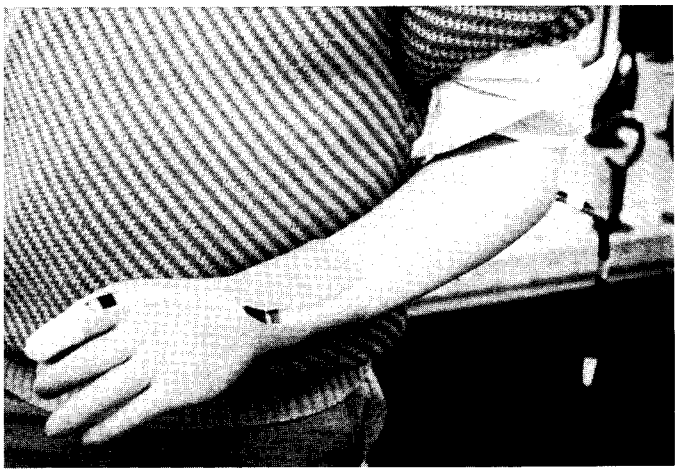

Figure 5.
The single degree of freedom SAMS hand in clinical application.

DISCUSSION

The ease with which individuals control prostheses show how adaptable human beings are. They can learn to use a range of non-natural inputs to assist in their control of the devices, for example, using the vibrations of the motor in the conventional myoelectric hand as it stalls to detect contact with the object (7). However, using the existing structures in the way most appropriate to the action is preferable.

Users of conventional, two-channel myoelectric hands can learn to apply muscular tension from opposing muscle groups to open and close their hands. The more natural occurance is to proportionately relate the flexion angle with the extensor muscle's tension, as with the voluntary opening aspect of the Southampton Hand.

That nonanthropomorphic methods work at all is an illustration of the adaptability of the individual. However, the average prosthesis user is not willing to invest a great deal of long-term effort in controlling a hand unless the benefits are substantially greater. Thus, a solution with minimal user effort is preferred.

In an experiment with limited numbers and time, it is difficult to draw firm conclusions about the efficacy of a particular system. The Southampton Hand has a number of factors that are different from the other devices. First is the proportional control, which has an important effect on the ease of control of the device (5,16); at that time no commercial devices offered such a facility, although in recent years it has become more common.

A second factor is the device's geometry. The more anthropomorphic hand can afford a wider range of grip postures, under any control philosophy. But the effectiveness of the control scheme is harder to demonstrate. A more direct comparison is possible with the single degree of freedom hand. A computer controlling the hand directly is "aware" when the fingers are touching each other or when the hand is touching an object. So the controller can respond to different situations, and the threshold for opening an empty hand can be set to require a small muscular tension, but releasing an object is made a much more deliberate act. Users of normal myoelectric hands do complain that it is too easy to inadvertently release a held object when the extension level is set low enough to make opening the hand less tiring.

As is apparent, the factors that dictate the acceptance or rejection of a prosthesis are many, varied, and unpredictable (33). One user may accept a prosthesis enthusiastically for precisely the same reason that another rejects it out of hand. Appearance, weight, action, ease of use, and cost all contribute to the decision. The Southampton Hand and its derivatives attempt to address the problems of ease of operation and appearance. Many of the other factors can only be addressed in a near-production, robust field version of the device.

Finally, as the hand is a semi-autonomous manipulator, it can be used in other fields beyond prosthetics. A robot manipulator need not look like a hand, but if it is to be used in a domestic environment it must hold objects found in that arena. Most such objects are designed to be held by human hands. Thus, a basically anthropomorphic shaped gripper is useful. In addition, if the users are individuals who have a restricted range of input, such as those with dysfunction brought on by a neuromuscular disorder (e.g., muscular dystrophy), then a simple high-level control channel for the gripper is an advantage, enabling an operator to make the most of his limited physical abilities. Such a philosophy will form a part of the Oxford Robot Assistant project (**Figure 6**) being conducted at the Oxford Orthopaedic Engineering Centre to develop different technologies to help people with special needs (34).

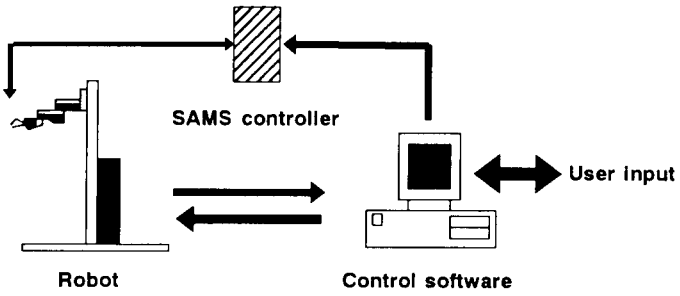

Figure 6.
Oxford Robot Assistant System, this combines the control of a SCARA robot with a semi-intelligent mobile base and a SAMS based gripper.

CONCLUSION

Dexterous manipulation of a range of objects is possible if a selection of special tools is used. A general manipulator must be adaptable in its geometry if it is to handle a wide range of objects. One such device is the Southampton Hand. The control is kept simple by divorcing the supervision of the device from the detailed management of the hand. In limited field experiments the combination of a multi-degree of freedom hand and hierarchical control showed improved performance in the

range of objects grasped and the tasks performed compared with the standard devices of the time.

This demonstrates the importance of the Southampton Adaptive Manipulation Scheme as a control technique for prostheses. It leaves an imperative to conduct more rigorous tests on the scheme in the field. The SAMS philosophy formed part of a collaboration funded by the European Community under the TIDE initiative, between centers in the UK and Italy, to develop a two degree of freedom intelligent myoelectric hand for clinical evaluation in both countries which began in 1993 (35).

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REFERENCES

1. Atkins DJ, Meier RH, eds. The comprehensive management of the upper-limb amputee. New York: Springer-Verlag, 1988.
2. Bennett WA. Limb Prosthetics. 6th ed. New York: Demos Publications, 1989.
3. LeBlanc M, Parker D, Nelson C. New designs for prosthetic prehensors. In: The 9th International Symposium on the Advances in External Control of Human Extremities, Dubrovnik, 1987:475–81.
4. Plettenburg RH, Cool JC. The WILMER cosmetic prosthetic prehensor for children. In: Proceedings of the International Society of Prosthetics and Orthotics, 7th World Congress, Chicago, 1992:62.
5. Milstien HG, Heger H, Hunter GA. Prosthetic use in adult upper-limb amputees: a comparison of body powered and electrically powered prostheses. Prosthet Orthot Int 1986:10:27–34.
6. Simpson DC, Kenworthy G. The design of a complete arm prosthesis. Biomed Eng February 1973:56–9.
7. Kyberd PJ, Mustapha N, Carnegie F, Chappell PH. Clinical experience with a hierarchically controlled myoelectric hand prosthesis with vibro-tactile feedback. Prosthet Orthot Int 1993:17(1):56–64.
8. Coghill-Smith HA. Data transfer via the skin. International Society of Prosthetics and Orthotics, UK annual scientific meeting, Nottingham, 1989.
9. Solomonow M, Lyman J, Freedy A. Fundamentals of electrotactile stimulation for sensory feedback from limb prosthetics. In: Proceedings of the Sixth International Symposium on the External Control of Human Extremities, Dubrovnik, 1978.
10. Gow DJ, Dick TD, Draper ERC, London IR, Smith P. The physiologically appropriate control of an electrically powered hand prosthesis. In: Proceedings of the International Society of Prosthetics and Orthotics, 4th World Congress, London, UK, 1983.
11. Gibbons DTO, Rain MD, Phillipe-Auguste JS. A multi-functional prosthesis employing extended physiological proprioception and programmed linkages. In: Proceedings of the first workshop on medical robotics, Ottawa, ON, 1988.
12. Meek SG, Jacobsen SC, Goulding PD. Extended physiological taction: design and evaluation of a proportional force feedback system. J Rehabil Res Dev 1989:26(3):53–62.
13. Doubler JA, Childress DS. An analysis of extended physiological proprioception as a prosthesis control technique. J Rehabil Res Dev 1984:21(1):5–18.
14. Hugh Steeper Ltd. Product catalogue 1992:L18236(4).
15. Otto Bock Orthopaedische Industry GmbH. MYOBOCK arm components 1992/93. Otto Bock, 1992.
16. Sears HH, Shaperman J. Proportional myoelectric hand control: an evaluation. Am J Phys Med Rehabil 1991:70(1):20–8.
17. Childress D. Historical aspects of powered limbs prostheses. Clin Prosthet Orthot 1985:9(1):2–13.
18. Bekey GA, Tomovic R, Zeljkovic I. Control architecture for the Belgrade USC hand. In: Venkataraman ST, Iberall T, eds. Dexterous robot hands. New York: Springer Verlag, 1990.
19. Yongsheng S, Li X, Feihong M. Development of a novel prosthetic prehensor. In: Proceedings of the International Society of Prosthetics and Orthotics, 7th World Congress, Chicago, 1992:190.
20. Todd RW. Adaptive control of a hand prosthesis (Dissertation). Southampton, UK: University of Southampton, 1969.
21. Lozach Y, Madon S, Hubbard S, Bush G. On evaluation of a multifunctional prosthesis. In: Proceedings of the International Society of Prosthetics and Orthotics, 7th World Congress, Chicago, 1992:185.
22. Baits JC, Todd RW, Nightingale JM. The feasibility of an adaptive control scheme for artificial prehension. In: Proceedings of a symposium at the Institute of Mechanical Engineers, 1968:183–3J:54–9.
23. Chappell PH, Kyberd PJ. The prehensile control of a hand prosthesis by a microcontroller. J Biomed Eng 1991:13:363–9.
24. Napier JR. The prehensile movements of the human hand. J Bone Joint Surg 1956:38B-4:902–13.
25. Rood ON. On the contraction of muscles, induced by contact with bodies in vibration. Am J Sci Arts 1860:25:499.
26. Edin B, Howe R, Westling G, Cutkowsky M. A physiological method for relaying frictional information to a human teleoperator. IEEE Trans Syst Man Cybern 1993:23(2):427–32.
27. Knox EH, Childress D, Heckathorne CW. Slip detection and automatic grasping for a prosthetic prehensor. In: Proceedings of the International Society of Prosthetics and Orthotics, 7th World Congress, Chicago, 1992:189.
28. Winkler W, Bierwirth W. The use of a gliding detector for the control of a prosthesis-hand. In: Proceedings of the International

Society of Prosthetics and Orthotics, 7th World Congress, Chicago, 1992:188.
29. Codd RD. Development and evaluation of adaptive control for a hand prosthesis (Dissertation). Southampton, UK: University of Southampton, 1975.
30. Kyberd PJ, Chappell PH. A force sensor for automatic manipulation based on the Hall effect. Measurement Sci Technol 1993:4:281–7.
31. Kyberd PJ, Chappell PH. Characterisation of an optical and acoustic touch and slip sensor for autonomous manipulation. Measurement Sci Technol 1992:3:969–75.
32. Chappell PH, Nightingale JM, Kyberd PJ, Barkhordar M. Control of a single degree of freedom artificial hand. J Biomed Eng 1987:9:273–7.
33. Senski T. A consumer's guide to "bionic arms." Br Med J 1980:12 Jul:126–7.
34. Kyberd PJ. User interface: the vital link. In: Proceedings of the Second UK Workshop in rehabilitation robotics. 1991:169–76.
35. Kyberd PJ, Tregidgo R, et al. The MARCUS intelligent hand prosthesis. In: Ballabio E, et al. eds. Rehabilitation technology: strategies for the European union. Amsterdam: IOS Press, 1993:98–102.

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Toward classification of dysphagic patients using biomechanical measurements

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Abstract—Identification of a patient at risk of aspiration is a major problem in the rehabilitation of the dysphagic patient. The present methods of diagnosis are based on clinical evaluation or videofluorography or fiberoptic endoscopic examination of swallowing (FEES). Recently, we developed biomechanical techniques for noninvasive quantitative assessment of the dysphagic patient. The purpose of the present investigation was to assess the clinical validity of the technique. In a double-blind study, both biomechanical test results and videofluorography (including bedside evaluation) results were used to independently classify the patients into four categories of risk for aspiration. Of the 36 patients studied, there was complete agreement between the biomechanical and clinical classifications in 21 patients. In 11 patients, the biomechanical technique overestimated the risk by one category, and underestimated the risk by one category in four patients. The biomechanical technique presents a useful tool for continued patient assessment; however, further studies are needed.

Key words: *accelerometry, aspiration, classification, diagnosis, dysphagia, noninvasive measurements, swallowing disorders.*

INTRODUCTION

Proper diagnosis of swallowing disorders (dysphagia) presents a continuing problem in the rehabilitation of patients with stroke and head injury, and of other patients with paralyzing neurological diseases. Dysphagia can occur as a result of lesions in certain cranial nerves, their nuclei, fiber tracts, or in the cortex. Aspiration is a major problem associated with dysphagia. Identification of the patient at risk of aspiration is important from the clinical viewpoint (1–3). The current clinical methods of diagnosis are qualitative and are based on clinical (bedside) evaluation and videofluorography examination (1–4). The videofluorographic examination (VFE) may involve radiation hazard and therefore may have a limitation as a diagnostic tool. Ultrasound has been suggested as a possible method of assessment (5,6). Investigators have used electroglottography as an assessment tool (7,8). Fiberoptic endoscopic examination of swallowing (FEES) has been used for clinical evaluation (9). However, there continues to be a need for developing noninvasive diagnostic techniques to identify the patient at risk of aspiration and to aid in the treatment of dysphagia. Recently, we identified and developed techniques to quantify noninvasively various biomechanical parameters that characterize the oral (10) and pharyngeal phases of swallowing (11). However, the question remains whether the patient at risk of aspiration can be identified and classified using the noninvasive biome-

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chanical measurements. The purpose of the present investigation is to address this question.

METHODS

Biomechanical Measurements of the Oral Phase

We have identified several biomechanical parameters involved in the oral phase of swallowing, and have developed techniques and instrumentation to noninvasively quantify these parameters (8). The biomechanical parameters are 1) lip closure (interface) pressure, 2) lip pulling (shear) force, 3) tongue thrust in the two lateral directions, 4) forward tongue thrust, and 5) swallow suction pressure.

The lip closure (interface) pressure was measured with an ultraminiature flat pressure transducer (Entran Devices Corp., Fairfield, NJ, Model EPL-125). The transducer, in a disposable wrapping paper, was placed between the lips at three different locations: right lateral, midline, and left lateral.

The lateral tongue thrust transducer was constructed by placing miniature strain gauges (Measurements Group Inc., Model EA-125-BZ-350) on a small plastic cantilever bar. Lateral forces on the bar exert a bending moment near the base which is detected by the strain gauge (**Figure 1**). The subject's head was placed in a head restraining system, as shown in **Figure 1**. For the measurement of forward tongue thrust, a cup was attached to the end of a small bar that ran through an ultraminiature load cell (**Figure 2a**).

A measure of the lip interface shear force was obtained by measuring the pulling force exerted by the lips on a flat spoon-like device. The lip interface pulling (shear) force transducer was constructed with a small plastic bar to which an ultraminiature load cell (Sensotech Inc., Columbus, OH, Model 121102) was attached (**Figure 2b**). The flat side of the transducer was placed between the lips, and the patient was asked to exert maximum pulling force with the lips. The pulling force was measured by the load cell. The subject's head was placed in the same head restraining system. Swallow suction pressure was measured with a miniature catheter connected to a hydraulic pressure transducer (Cobe, Inc., Lakewood, CO). Measurements obtained from dysphagic patients were signifi-

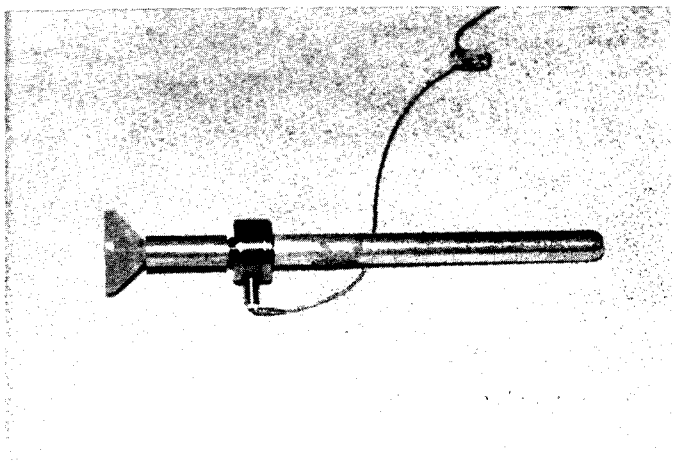

Figure 2a.
The forward tongue thrust transducer consists of a small cup-like dish attached to a miniature plastic bar with a miniature load cell (force transducer). The patient is asked to push the cup with his tongue while the device is held stationary.

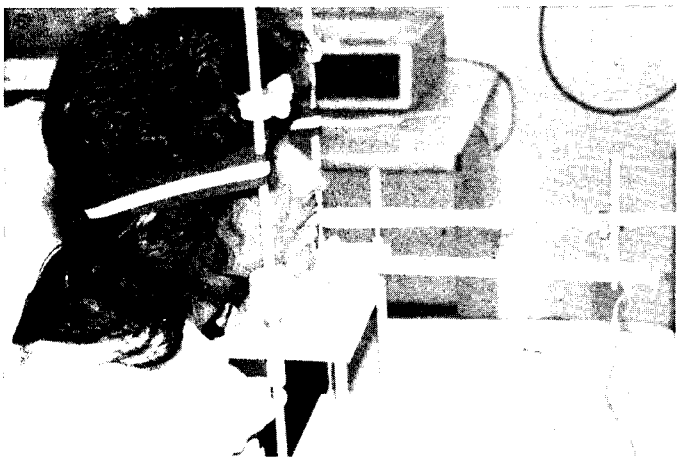

Figure 1.
Lateral tongue thrust measurement system consists of a strain-gauged miniature cantilever plastic bar. The patient's head is positioned in a jig during the measurement.

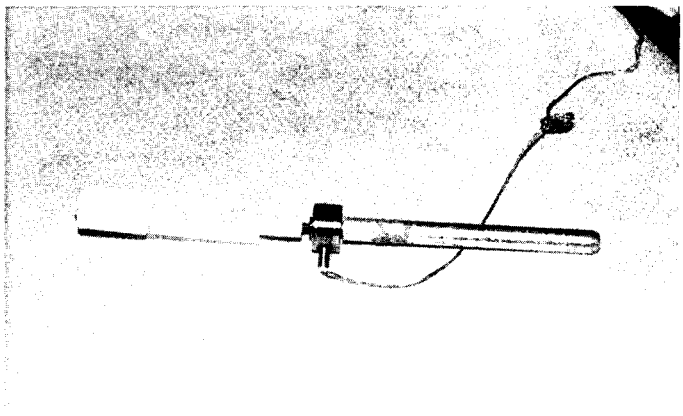

Figure 2b.
The lip interface shear force transducer consists of a small flat plastic bar attached to a miniature bar through a load cell (force transducer).

cantly lower in magnitude when compared to nondysphagic subjects (10).

Biomechanical Measurement of the Pharyngeal Phase

We have developed the accelerometry technique to noninvasively assess the swallowing mechanism (11). We placed two ultraminiature accelerometers (Kulite Semiconductor Inc, Ridgefield, NJ, Model GY-125-10) on the skin at the throat (**Figure 3**). Accelerometer 1 was placed at the midline at the level of the thyroid cartilage and accelerometer 2 at the midline at the level of the cricoid cartilage. Simultaneously, we measured the swallow suction pressure with a catheter, with the tip placed on the midline toward the posterior aspect of the tongue, and connected to a hydraulic pressure transducer (Cobe Inc.).

In nondysphagic individuals, swallowing gave rise to a characteristic acceleration pattern (**Figure 4**). The peak acceleration during swallowing ranged from 1 to 3 g. The swallow pressure was biphasic. There was no latency (time lag) between the appearance of the pressure wave and appearance of the acceleration wave characteristic of swallowing. Moreover, only one attempt was required to produce a swallow.

In dysphagic patients, the acceleration pattern typical of swallowing was either absent (**Figure 5**) or delayed (11). In those patients who could trigger a swallow, significant lag times were found between the pressure waveform and the acceleration waveform. In patients who could trigger a swallow, several attempts were required to produce the response. The amplitude of peak acceleration measured in dysphagic patients ranged from 0 to 0.5 g.

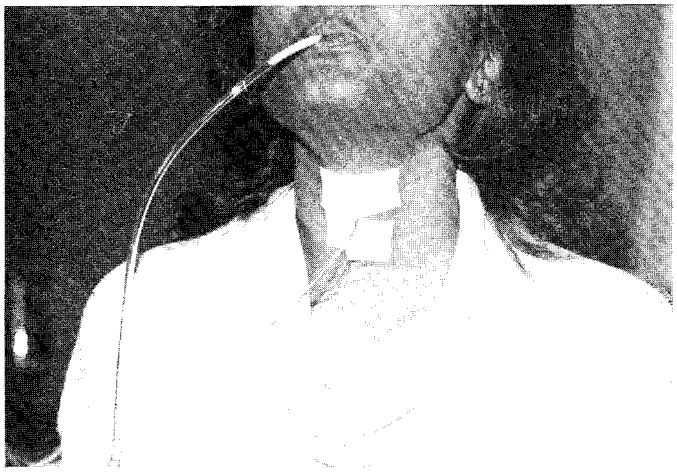

Figure 3.
The pharyngeal phase is assessed by attaching two ultraminiature accelerometers on the skin over the throat over thyroid and cricoid cartilages. A catheter is placed midline in the mouth toward the posterior aspect of the tongue to measure swallow pressure.

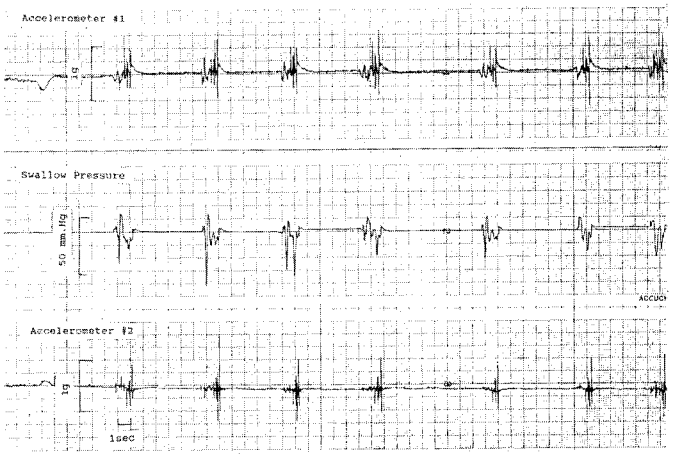

Figure 4.
Acceleration and swallow pressure measurements obtained from a nondysphagic subject. Top and bottom accelerometers drive the top and bottom channels; the middle channel reflects swallow pressure. Swallow pressure is biphasic (changes direction) and the characteristic nondysphagic acceleration pattern is developed. There is no latency between the swallow pressure and acceleration responses.

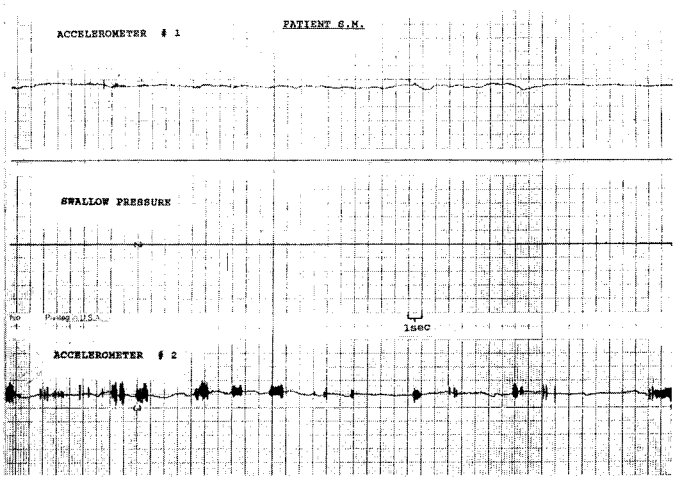

Figure 5.
Acceleration and swallow pressure response obtained from a patient.

Classification of the Risk for Aspiration Using the Biomechanical Measurements

In 36 physician-referred patients from whom consent was obtained, we made biomechanical measurements of the oral and pharyngeal phases within 7 days of the clinical

bedside evaluation and VFE in a double-blind study. The biomechanical results and VFE and clinical bedside evaluation results were independently classified into four categories of risk for aspiration: 1) normal (no risk), 2) mild risk, 3) moderate risk, and 4) severe risk. The VFE and clinical classification was made by a single speech pathologist. The biomechanical classification was made by a biomedical engineer. Both classifications were compared at the end of the study. The investigators conducting the biomechanical evaluation did not know the patient etiology or clinical classification. Similarly, the clinical evaluator did not have the biomechanical measurements or classifications. Also, at the end of the study, the biomedical engineer examined the VFE records and compared them with certain features in the biomechanical recordings.

First, the oral biomechanical measurements were classified into four categories. Then, the pharyngeal biomechanical (accelerometry) measurements were classified into four categories, and finally, an overall biomechanical classification was made of the combined oral and pharyngeal phases.

The oral biomechanical measurements were classified as follows:

Category	Criteria
a) normal	near normal lip closure pressure (**LP**) AND near normal tongue thrust (**TT**) in all directions: [**LP** ≥ 70 mmHg] **AND/OR** [**TT** ≥ 250 g]
b) mild	low tongue thrust AND/OR low lip closure pressure: [15 mmHg < **LP** < 70 mmHg] **AND/OR** [125 g < **TT** < 250 g]
c) moderate	moderately lower tongue thrust AND/OR moderately lower lip closure pressure: [5 mmHg < **LP** < 15 mmHg] **AND/OR** [20 g < **TT** ≤ 125 g]
d) severe	very low lip closure pressure AND/OR very low tongue thrust: lip closure pressure below 5 mmHg AND/OR tongue thrust below 20 g. [**LP** ≤ 5 mmHg] **AND/OR** [**TT** < 20 g]

The pharyngeal phase biomechanical measurements were classified as follows:

Category	Criteria
a) normal	normal acceleration pattern (**AP**) AND near normal acceleration magnitudes (**AM**) AND near normal suction pressure (**SP**) One number of attempts to swallow (**NA**): [Normal **AP**] **AND** [**AM** ≥ 0.7 g] **AND** [**SP** ≥ 40 mmHg] **AND** [**NA** = 1]
b) mild	[Normal **AP**] **AND** [**NA** < 2] **AND** [0.5 g < **AM** < 0.7 g) **AND/OR** (**SP** < 40 mmHg)]
c) moderate	[(Slightly distorted **AP**) **AND/OR** (**2** < **NA** < 4)] **OR** [(0.2 g < **AM** < 0.5 g) **AND/OR** (**SP** < 30 mmHg)]
d) severe	very low acceleration magnitude (below 0.2 g) or very low swallow pressure (below 10 mmHg) OR more than four attempts to swallow OR significantly distorted acceleration pattern [(Significantly distorted **AP**) **AND/OR** (**NA** ≥ 4)] **OR** [(**AM** < 0.2 g)] **OR** [(**SP** < 10 mmHg) **AND** (**SP** Monophasic)]

Each patient was then assigned an overall biomechanical rating based on the biomechanical classification in the oral and pharyngeal phases. The overall rating was given as described below:

Category	Criteria
a) normal	normal oral phase AND normal pharyngeal phase
b) mild	normal oral phase AND mild pharyngeal phase OR mild oral phase AND normal pharyngeal phase OR mild oral phase or mild pharyngeal phase

c) moderate	moderate oral phase AND mild or normal pharyngeal phase OR moderate pharyngeal phase AND mild or normal oral phase OR moderate oral phase AND moderate pharyngeal phase OR severe oral phase AND mild pharyngeal phase
d) severe	severe oral phase AND moderate pharyngeal phase OR severe pharyngeal phase

Clinical and VFE Assessment

The biomechanical classification was correlated with the classification assigned by the speech pathologist after the double-blind study. The speech pathologist's classification was based on findings of the clinical (bedside) evaluation and VFE.

The clinical evaluation of the pharyngeal phase included assessment of velar function in terms of elevation, asymmetry, gag, and nasality, and assessment of the laryngeal function in terms of hoarseness, breathiness, gurgling, pitch and loudness of sound, and voluntary and reflective cough strength. Clinical evaluation of the oral phase included evaluation of labial and lingual function in terms of range of motion, coordination, and strength.

The VFE was conducted with thick liquid and pureed consistency. The patient was in an upright seated position. The examination involved observation of the oral phase in terms of tongue motion, residue in oral cavity, oral transit time, and premature pharyngeal entry. The assessment of the pharyngeal phase involved laryngeal elevation, pharyngeal transit time, laryngeal penetration, and residue in valleculae, pyriform sinus, and laryngeal vestibule. Each parameter was judged normal, mild, moderate, or severe. An overall classification was assigned for each patient based on all these parameters as a part of the hospital protocol. This classification was compared with the biomechanical classification at the completion of the double-blind study.

Data Analysis

A non-parametric statistical test (Wilcoxon test) was performed to study the statistical significance in the differences between the biomechanical and clinical classifications based on VFE and clinical bedside examination.

Observations

After the comparison of the biomechanical and clinical classifications, one of the authors examined the VFE records and made observations relating the events of the VFE with the following characteristics of the pharyngeal biomechanical recordings: acceleration pattern, acceleration magnitude, high frequency noise (blurring in the record) in the acceleration recording, and magnitude of swallow suction pressure.

RESULTS

In 21 of the total of 36 patients, there was complete agreement between the biomechanical classification and VFE and clinical classification of risk for aspiration (**Table 1**). In 11 patients, the biomechanical classification overestimated the risk by one category. In four patients, the biomechanical classification underestimated the risk by one category. Wilcoxon test (a non-parametric test) did not indicate any significant statistical difference between the two methods.

The acceleration response of patient **M.W.** with a mild pharyngeal phase risk of aspiration is shown in **Figure 6**. The peak acceleration magnitude was 1.17 g from accelerometer 1 and 2.3 g from accelerometer 2. The swallow suction pressure was 35 mmHg. The patient had

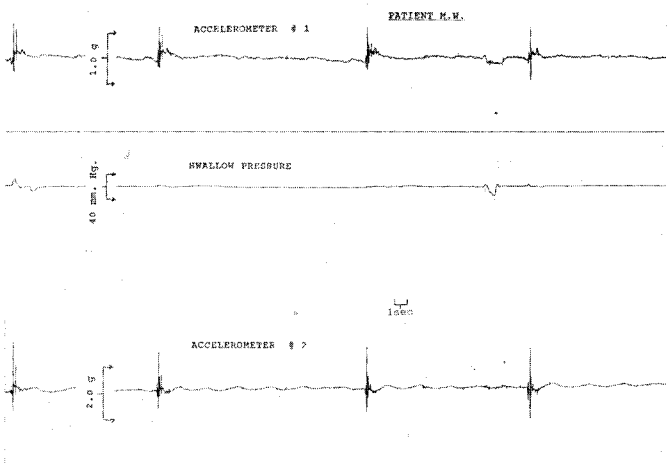

Figure 6.
Acceleration and swallow pressure recorded from dysphagic subject **M.W.** with mild risk of aspiration.

Table 1:
Correlation of biomechanical and clinical risk ratings.

PATIENT	ORAL PHASE Biomech	PHARYNG PHASE Biomech	OVERALL BIOMECH RATING	VFE & CLINICAL EVALUATION	PATIENT ETIOLOGY	Age
M.P.	Moderate	Moderate	Moderate	Moderate	RCVA	71
M.V.	Mild	Moderate	Moderate	Severe	RCVA	75
M.M.	Moderate	Moderate	Moderate	Severe	Guillain Barré Syndrome	37
P.C.	Mild	Moderate	Moderate	Mild	LCVA, diabetes	72
A.F.	Moderate	Severe	Severe	Severe	Tubercular Meningitis	52
G.M.	Severe	Severe	Severe	Moderate	RCVA	90
R.R.	Mild	Moderate	Moderate	Moderate	CVA	67
G.J.	Severe	Severe	Severe	Severe	LCVA	77
E.M.	Moderate	Mild	Moderate	Mild	LCVA	65
H.P.	Moderate	Mild	Moderate	Moderate	LCVA	73
J.G.	Mild	Moderate	Moderate	Moderate	CHI, anoxia	49
D.A.	Moderate	Mild	Moderate	Moderate	LCVA	76
G.H.	Moderate	Mild	Moderate	Moderate	LCVA	79
D.S.	Mild	Severe	Severe	Severe	CHI	19
E.P.	Severe	Severe	Severe	Severe	LCVA	69
B.S.	Moderate	Severe	Severe	Moderate	RCVA	67
M.W.	Severe	Mild	Moderate	Mild	Cerebral Infarction	75
F.S.	Moderate	Mild	Moderate	Moderate	Mult. Laminar Infarction	68
H.J.	Moderate	Mild	Moderate	Moderate	LCVA	72
G.M.	Mild	Moderate	Moderate	Moderate	RCVA	90
W.C.	Severe	Severe	Severe	Severe	Astrosystoma Brainstem	43
V.F.	Severe	Severe	Severe	Moderate	LCVA	82
S.G.	Moderate	Severe	Severe	Severe	CHI	29
M.C.	Moderate	Mild	Moderate	Moderate	LCVA	76
D.M.	Severe	Severe	Severe	Severe	RCVA	83
G.W.	Moderate	Severe	Severe	Severe	Bilateral CVA	73
T.N.	Mild	Moderate	Moderate	Mild	Bilateral CVA	45
D.F.	Moderate	Severe	Severe	Moderate	Anoxia	62
J.V.	Moderate	Mild	Moderate	Mild	LCVA	49
C.G.	Moderate	Severe	Severe	Severe	Bilateral CVA	50
E.S.	Mild	Severe	Severe	Moderate	RCVA	81
J.S.	Mild	Mild	Mild	Moderate	LCVA, Lacunar Infarction	84
G.W.	Moderate	Severe	Severe	Moderate	Bilateral CVA	73
M.T.	Mild	Moderate	Moderate	Moderate	Parkinson's Disease	76
D.M.	Moderate	Severe	Severe	Severe	Subarchanoid Hemorrhage	49
H.G.	Mild	Moderate	Moderate	Severe	RCVA	73

PHARANG = pharangeal; VFE = videoflouroscopic examinaton; CVA = cerebrovascular accident; RCVA = right CVA; LCVA = left CVA; CHI = closed head injury

a lip closure pressure of only 5 mmHg, and a lip pulling force of only 33 g. The patient was unable to perform the tongue thrust and had a 0 g of lateral tongue thrust. Biomechanically, the patient was classified as having a mild risk for the pharyngeal phase and a severe risk for the oral phase. He was given an overall biomechanical rating of moderate. His VFE and clinical bedside evaluation revealed that he had a mild pharyngeal phase dysphagia. A delayed triggering of the pharyngeal phase was noted primarily due to reduced initiation in the oral phase. He had moderate to severe oral phase dysphagia clinically. However, the overall clinical rating was mild. **Figures 7** and **8** show the acceleration patterns of two patients (**D.M.** and **W.C.**) who were classified at significant risk by both biomechanical and VFE and clinical examinations.

Figure 9 shows the acceleration and pressure measurements from patient **H.J.** at moderate risk for aspiration. The peak acceleration magnitudes were 2 g for accelerometer 1 and 1.25 g for accelerometer 2. He had a swallow pressure of 33.2 mmHg. Although the magnitudes of ac-

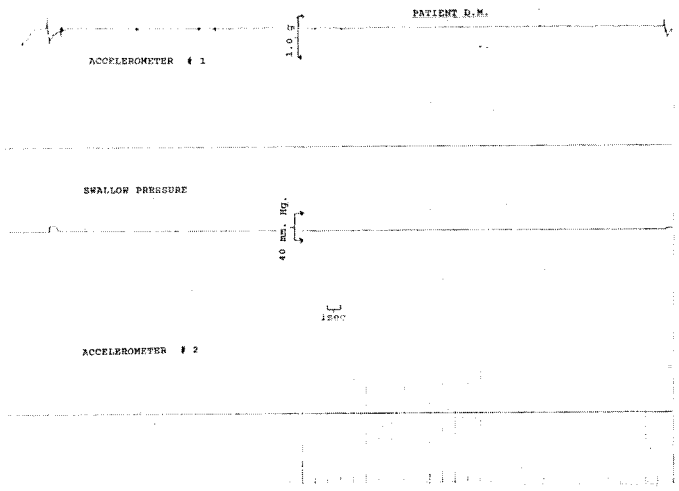

Figure 7.
Acceleration and swallow pressure measurements obtained from dysphagic patient **D.M.** with significant risk for aspiration.

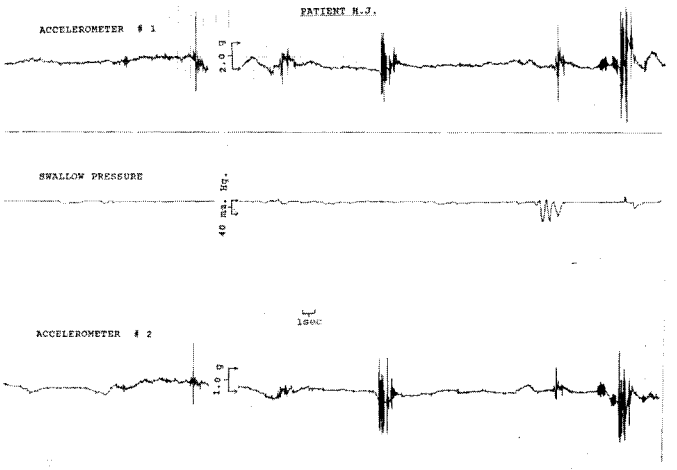

Figure 9.
Acceleration and swallow pressure response obtained from dysphagic patient **H.J.** with moderate risk for aspiration.

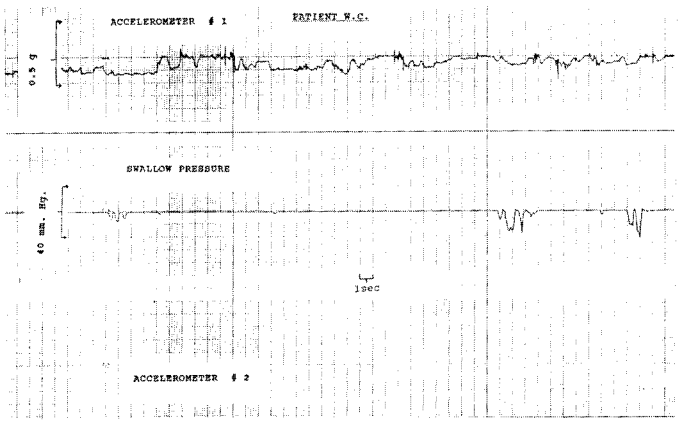

Figure 8.
Acceleration and swallow pressure measurements obtained from dysphagic patient **W.C.** with significant risk for aspiration.

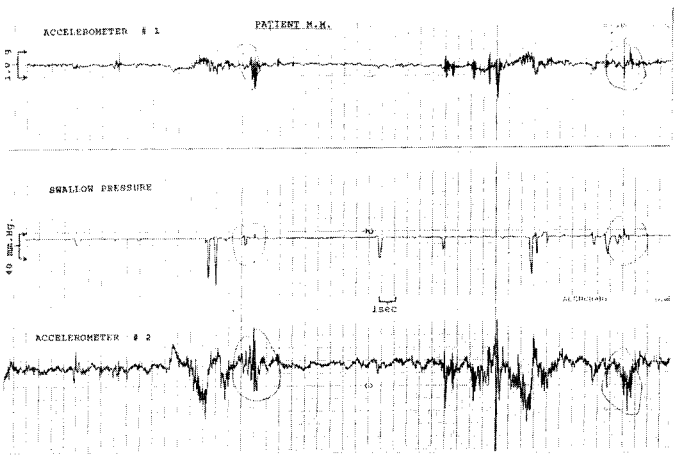

Figure 10.
Acceleration and swallow pressure response obtained from dysphagic patient **M.M.** with significant risk for aspiration.

celeration were normal, the acceleration pattern had a slight high frequency blurring. His oral measurements were not obtained as he had dentures which interfered with the measurements. The patient was assigned an overall biomechanical rating of moderate risk. Clinically, the patient had a moderate oral phase dysphagia. He had mild to moderate amounts of residue in the valleculae and demonstrated trace residual pooling in the pyriform sinuses. The patient was assigned an overall clinical rating of moderate risk.

The acceleration and swallow suction pressure patterns of patient **M.M.**, who is at severe risk of aspiration, is shown in **Figure 10**. The VFE revealed that the patient had severe pharyngeal phase dysphagia with a moderate to marked residue in the vallecular spaces which she was unable to clear with repeated swallows. The biomechanical evaluation revealed that the acceleration pattern had blurring. Although the acceleration pattern was distorted, one swallow was noted in five minutes. Since the swallow was noted, the patient was given a biomechanical classification of moderate risk. Her clinical classification was severe risk.

Several observations were made by the biomedical engineer when examining the VFE records after the completion of the double-blind study. Although the following

observations were made qualitatively, further systematic investigation is needed. In cases of patients who had food stasis or food residue in the vallecular spaces as confirmed by the VFE, the acceleration response showed slight high frequency blurring (high frequency noise in the acceleration signal). For instance, patient **F.S.** had food residue and also had laryngeal penetration as observed by the videofluorography, and had high frequency noise (blurring in the record) in the acceleration response shown in **Figure 11**. This case can be contrasted with the acceleration recording from patient **M.W.** who did not have any food stasis and the acceleration record is free of high frequency noise (**Figure 6**). The magnitude of the characteristic acceleration was small in patients who had poor laryngeal elevation. The characteristic acceleration pattern was absent in patients who had little or no laryngeal adduction. **Figures 7** and **8** show the acceleration patterns from two such patients. Pumping action was observed in the swallow pressure response in patients who required a large number of attempts to swallow (e.g., patient **G.W.** in **Figure 12**). The acceleration signal from a dysphagic patient with Parkinson's disease (**M.T.**) contained low frequency oscillations (**Figure 13**).

DISCUSSION

The present results have demonstrated the potential of the biomechanical measurements in noninvasive assessment of the dysphagic patient, and in the identification of the patient at risk of aspiration. The biomechanical technique presents the clinician with a noninvasive tool to demonstrate the clinical improvements for continued patient assessment in terms of the biomechanical parameters. Moreover, the biomechanical method can be used more frequently to aid in the identification of patients who need a VFE.

Currently, there is no unique method of classification of risk for aspiration, and the clinical classification itself is somewhat subjective and could vary from clinician to clinician. In spite of this, there was complete agreement in

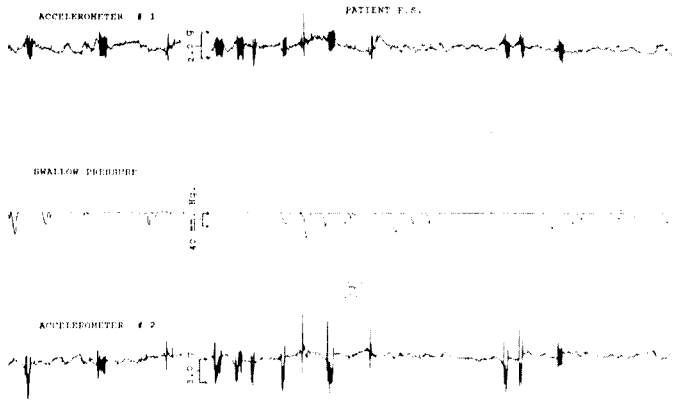

Figure 11.
Acceleration and swallow pressure measurements obtained from dysphagic patient **F.S.** who had pharyngeal penetration.

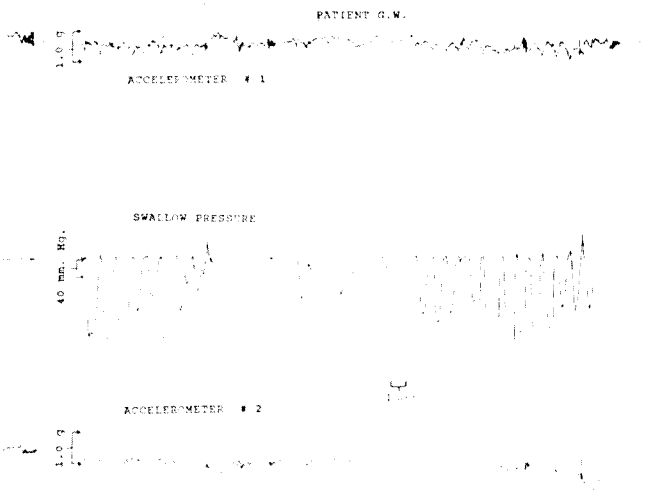

Figure 12.
Acceleration and swallow pressure measurements obtained from dysphagic patient **G.W.** who required many attempts to swallow. The biomechanical classification was significant risk and the clinical classification was moderate risk for aspiration.

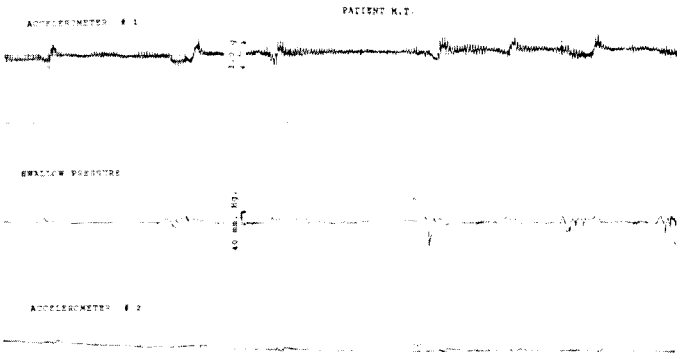

Figure 13.
Acceleration and swallow pressure measurements obtained from a dysphagic patient with Parkinson's disease.

21 of 36 patients and the biomechanical technique overestimated the risk in 11 subjects and underestimated the risk in 4 subjects by one category. The disagreement could be attributed to several factors. First, the positioning of the patient may be an important factor. In the biomechanical examination, the patient is seated vertically. In the VFE, the patient may be examined in a number of positions (12). A semi-erect position can assist the disordered patient in compensating for the dysphagia (12). Second, the biomechanical method involved dry swallow. In the VFE, the patient is given food boluses of various consistencies. Viscosity was progressively changed to test the patient's tolerance depending upon other conditions, such as etiology, age, or history. Third, in the biomechanical measurement, the patient is given a 5-minute duration for the pharyngeal examination during which the patient may or may not swallow. In the VFE, the patient is given food boluses of various consistencies, and the speech pathologist and/or the radiologist wait(s) for the patient to initiate a swallow. The duration of the wait is left to the discretion of the radiologist. Finally, the condition of the patient might have changed slightly between the VFE and biomechanical measurement.

The noninvasive biomechanical measurement procedure required from 20 to 30 minutes. The present recording instrumentation is not portable; therefore, patients who could not be readily transported to the laboratory were excluded from the study. Also excluded were those patients who could not follow instructions.

Success of the biomechanical method depends on the cooperation, mood, and alertness of the patient. Cognition is an important factor. For instance, a head injury patient may have a very short attention span or be irritable and noncooperative. Measurements obtained from such a patient may not be as comprehensive or accurate as desired.

The high frequency noise observed in some patients might be occurring due to the presence of food or food penetration and should be further investigated. Also, smaller magnitudes of acceleration observed in patients with poor laryngeal elevation, and distorted or absent acceleration patterns observed in some patients with little or no laryngeal adduction, were only observations in some patients and should be further investigated systematically.

In the present investigation, only 36 subjects were used. With the limited number of subjects, it is difficult to confirm in which categories the biomechanical method is most effective in the assessment of the risk. Also, the present investigation with a limited number of patients cannot conclusively reveal which parameter is most indicative of the risk for aspiration. Parameters of the pharyngeal phase may be very important. However, oral phase parameters may also be important. For instance, the lateral tongue thrust indicates the ability for tongue lateralization. Control of the tongue is important to prevent premature pharyngeal entry. Further investigations are being undertaken to unravel the importance of each of the parameters.

Arbitrary rules were used in the present study to classify the biomechanical measurements into various risk categories. The parameters separating the categories (mild, moderate, etc.) were chosen based on the experience from our previous studies (10–12). In reality, we have an expert system based on limited previous data. With continued studies with a large number of patients, this classification procedure can be improved. Nevertheless, the present investigation represents the first step in using the biomechanical measurements for patient classification. Perhaps a computerized neural network model can be developed to classify patients based on the biomechanical measurements. Such models and measurements are being undertaken.

Compact, portable instrumentation is needed to quantify the swallowing mechanism and to detect aspiration at the bedside. With further refinement, the biomechanical techniques facilitate the development of such instrumentation. However, automatic pattern recognition techniques have to be developed to automatically classify the acceleration patterns such that the measurement and interpretation can be made by the clinician. Although two accelerometers were used in the present study, only one accelerometer placed at the thyroid cartilage is sufficient.

In current clinical practice, the swallowing mechanism is evaluated by clinical evaluations and the VFE. The clinical evaluation involves visual observation of the oromotor functioning in terms of the strength, coordination, range of motion, and functional tasks in swallowing. Laryngeal elevation is also examined as a part of the bedside evaluation. Vocal quality is examined for signs of gurgly quality which would indicate laryngeal and/or pharyngeal food residues. The VFE is generally used to confirm the clinical evaluation in those patients in whom there is a risk of aspiration or penetration. This examination involves the administration of food boluses of various consistencies mixed with barium. The patient is asked to swallow, and each swallow is carefully examined frame-by-frame to detect the movement of the bolus in relation to the structures.

In clinical practice, prescription and administration of proper food depends on the clinician's determination of the

dysphagic patient's risk for aspiration, his nutrition, and medical condition. The number of risk categories used varies from hospital to hospital. The actual classification is subjective as the clinician takes into account the VFE, the bedside evaluation, patient etiology, age, and other conditions. Also, the determination of the actual risk may require patient follow-up.

Since the videofluorographic technique may cause radiation exposure, repeated examination on a daily basis cannot be performed. Fiberoptic endoscopic examination of swallowing (FEES) developed by Langamore et al.(9) is a direct assessment tool and can be performed on a daily basis, but is rather invasive. Ultrasound has been suggested as a diagnostic tool for evaluation of the oral phase. Electroglottography (EGG), a method based on impedance change, has been suggested recently for the detection of the pharyngeal stage (7). However, the latter two methods have not been used in clinical practice.

The biomechanical method, therefore, presents a useful tool for comprehensive, quantitative, and noninvasive assessment of the dysphagic patient, and for identification of the patient at risk of aspiration. This technique complements other techniques such as the VFE, FEES, ultrasound, and EGG. Before the method is adopted for clinical use, studies are needed correlating the events of the biomechanical recordings with the motions of various structures involved in the pharyngeal phase of swallowing. Moreover, the biomechanical technique is not indicated in patients with poor cognition and in patients on whom the transducers cannot be placed (e.g., patients with neck braces or tracheal tubes). Also, the tongue thrust measurement requires placing the head in a jig to restrain the jaw, head, and neck motion. Only the pharyngeal phase measurements can be obtained in such cases. The biomechanical method may be useful in patients who cannot tolerate the risk of aspirating very small quantities of food. Nevertheless, the method complements the other existing techniques and should be investigated further. A study on a larger population of patients is necessary.

CONCLUSIONS

The biomechanical measurement technique presents a potential tool for noninvasive identification of the dysphagic patient at risk of aspiration; however, further studies on a larger population of patients are necessary before the technique can be used in the clinic. This technique may be combined with clinical bedside evaluation to determine the need for the VFE. Further refinements in terms of automation might be necessary to facilitate clinical application.

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REFERENCES

1. Logemann JA. Evaluation and treatment of swallowing disorders. Boston: Little Brown and Co., 1983.
2. McIntosh J, Martin G, Sacchett C. Recognition of the diverse and complex nature of neuromuscular swallowing disorders: a basis for treatment. Clin Rehabil 1987:1:39–45.
3. Veis SL, Logeman JA. Swallowing disorders in persons with cerebrovascular accident. Arch Phys Med Rehabil 1985:66:372–5.
4. Logemann JA. Manual for the videofluorographic study of swallowing. Boston: Little Brown and Co., 1985.
5. Sterns D. Ultrasound swallow. Br Med J 1978:2:1789–90.
6. Sonies BC. Instrumental procedure for dysphagia diagnosis. Semin Speech Lang 1991:12:185–92.
7. Sorin R, McClean MD. Electro-glottographic examination of swallowing. Arch Phys Med Rehabil 1987:68:232–5.
8. Perlman AL, Grayhack JP. Use of electroglottograph for measurement of temporal aspects of swallow: preliminary observation. Dysphagia 1991:6:88–93.
9. Langamore SE, Schatz K, Olson NO. Endoscopic and videofluoroscopic evaluation of swallowing and aspiration. Ann Otol Rhinol Laryngol 1991:100:678–81.
10. Reddy NP, Costarella BR, Grotz RC, Canilang EP. Biomechanical measurements to characterize the oral phase of dysphagia. IEEE Trans Biomed Eng 1990:37:392–7.
11. Reddy NP, Canilang EP, Casterline J et al. Noninvasive acceleration measurements to characterize the pharyngeal phase of swallowing. J Biomed Eng 1991:13:379–83.
12. Jones B, Donner MW. How I do it: examination of the patient with dysphagia. Dysphagia 1989:4:162–72.

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Voice output reader for displays on video cassette recorders and other domestic products

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Abstract—The increasing use of electronic displays in domestic products can pose great problems for blind people wishing to use microwave ovens, video cassette recorders, hi-fi systems, etc. One method of trying to solve these problems is to provide a handheld "display reader" that can translate the alphanumeric and symbolic information presented on the display into speech. A prototype system has been developed and evaluated. User requirements for this device and the problems of the image sensor, the image processing method, and the user interaction with the system are also discussed.

Key words: *assistive technology, blind, computers, consumer electronics, image processing, speech synthesis, visual impairment.*

INTRODUCTION

The two main areas identified as posing difficulties for blind people (1), are mobility (getting from A to B) and reading. Although the latter was originally perceived as being concerned with access to printed material, this now includes access to information generally, and, in particular, information stored and presented in systems employing computer technology. Considerable efforts have been made to enable blind people to access computer systems, and a significant area of recent interest has been providing access to computer systems employing a graphical user interface (GUI). These have resulted in a number of solutions predominantly for IBM-compatible equipment (2,3); it should be noted, however, that the first commercially available GUI access system was for the Macintosh [outSpoken from Berkeley Systems (4)].

In parallel with developments in desktop computing, there also has been an increasing tendency to use electronic displays and keypads in products such as video cassette recorders (VCRs), washing machines, and microwave ovens. As these displays become more commonplace, they can pose a serious obstacle to blind people operating these devices, which traditionally could be modified by adding tactile markings to the dials and switches of the products. This difficulty was highlighted recently by the World Blind Union Research Committee, who listed, among their priorities for technical research and development for visually disabled persons, a liquid crystal display (LCD) reader for numeric displays with speech output for less than 100 dollars (5). The work reported here has been undertaken to address this need by producing a prototype system that is suitable for further development to produce a small, handheld device with a self-contained power supply and capable of speaking the information found on typical consumer products with electronic displays.

Accessing Domestic Products

Toward the end of 1992 a survey of major United Kingdom retailers was undertaken to determine the state of the market in domestic products with electronic displays. The products chosen for the survey ranged from VCRs and microwave ovens to digital clocks and calculators. Displays on personal computers and organizers were not considered. Of particular concern were "leading-edge" products; these were examined with the objective of improving our understanding of both current and future trends

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in display technology and techniques. Displays were classified in terms of the underlying display technology, how the technology was used to present different data items on the display, and the structuring of the information on the displays.

Types of Display Technology

Three different types of display technology were found: 1) liquid crystal displays (LCDs); 2) light-emitting diodes (LEDs); and 3) gas plasma displays (GPDs).

LCDs are low-contrast, dark-on-light displays and are gradually becoming less common on domestic products in general, due to their limited viewing angles and relatively low contrast. However, they are common on battery powered devices, due to their low power consumption.

LEDs are bright and clear, light-on-dark displays, generally found on many types of domestic equipment. They have generally been superseded by GPDs, but are still found generally on simple, low-cost displays, such as alarm clocks.

GPDs were found to be the most common form of display used. Like LEDs they are bright and clear, but can be constructed to provide higher resolution, and therefore to produce more detailed displays.

Presentation of Information

The basic display technology can be used to present information in a number of ways. These include: 1) 7-segment displays; 2) 16-segment displays; 3) display symbols; and, 4) bit-mapped displays.

1. Seven-segment displays have the display segments arranged in such a way as to form numerals and a small subset of the alphabet.
2. Sixteen-segment displays are similar to seven segment displays but have an extra nine segments, which permit a better representation and a wider range of alphanumeric characters. These were found to be the most common displays in use.
3. In display symbols technology, a part of the display is constructed to present a particular symbol that can be "on" or "off." For example, the dots used to indicate a.m. or p.m. on some digital clocks, or the symbols that indicate the state of a VCR (i.e., play, pause, etc.). These symbols, referred to as "static elements" are found on almost all displays.
4. In bit-mapped displays a wide variety of symbols may be represented from a matrix of pixels. At the moment these are only present on top-of-the-range products, but it is anticipated that they will increase in usage over the next few years. It is important to note that this type of display has significantly different characteristics from the other display types in that the symbols can move in the display area.

Structure of Information

Figure 1 shows a display that is typical of those found on VCRs. This figure illustrates some of the terms used to describe and structure information found on displays. The display is considered to be made up of a number of *fields*, which are in turn made up of *elements*, which in turn are made up from *cells*. Cells are the fundamental units on the display that can be independently turned on or off. In the example, cells would include the recording icon, the "SUN" area, or any of the individual segments of the digits. Cells are grouped together to form elements. Elements may be alphanumeric or static symbols, which are made up of one or more cells. In the example above, elements made up of single cells include the recording icon and the "SUN" area; elements made up of multiple cells include the individual seven segment digits. Fields are sets of elements that, as a whole, form the fundamental units of information to be spoken. In the example above, four distinct fields can be identified as follows:

- a time field, consisting of four numeric elements and a separator (":")
- a tape counter field, consisting of five numeric elements
- a day-of-the-week field, consisting of seven static elements ("SUN," "MON," . . . "SAT"), which under most circumstances, would have a single element lit.

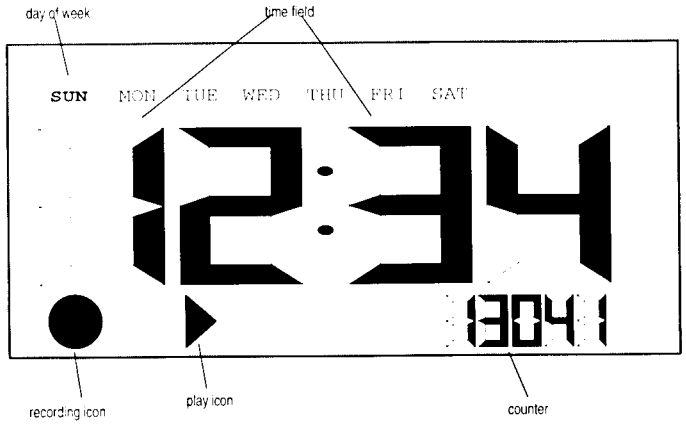

Figure 1.
An example of a display on a typical VCR.

- a state of system mechanism field, (i.e., a symbol representing RECORD, PLAY, FAST FORWARD, REWIND, PAUSE, etc.). In this case, between zero and three elements would be displayed. For example: zero in the case of the tape not being played; one for the tape being rewound; two in the case of "RECORD" and "PLAY"; and three in the case of a paused recording.

It should be noted that at any one time a user is typically interacting with one of the fields. In addition, on some systems information is scattered around the product (e.g., the "recording on" light on some VCRs is well away from the other information), or when information is not displayed on the product (e.g., with "on screen" VCR information). The work discussed here only deals with systems that indicate the system state in a single display panel.

Issues in Developing a Display Reader System

To address the difficulties of accessing domestic products, the authors propose a low-cost display reader. The system would be a small, handheld device with a self-contained power supply. The user would place the system in front of a display, possibly locating it in some form of jig. The system would then present information appropriate to the current display and system status to the user with synthetic speech. Users would have selective control over the information relayed to them by a set of user-interface buttons. The display reader system, attached to an electronic system with the relevant type of display, is shown in **Figure 2**.

In developing such a system the authors had to address two issues: the usability of the device and the technological issues pertaining to the "reading" of the display.

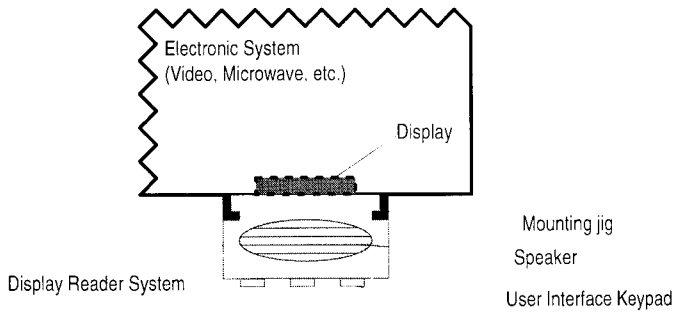

Figure 2.
Plan view of the display reader system attached to an electronic system.

The usability issue breaks down into three areas: device portability, the mechanical properties of the device, and the user interface.

A portable device must be of reasonably small size and also be cordless (i.e., it must be battery powered and hence the system design must minimize power consumption). To satisfy these requirements the system's chip count must be minimized, probably by using standard-cell application specific integrated circuit (ASIC) technology. If a sufficient volume of sales (i.e., in the order of hundreds) could be generated, this technology would significantly reduce the cost of the system.

The mechanical properties of the device impact greatly on its usability: users must be able to accurately position the device in front of the product's display and maintain this position for the period in which the product is in use. This prototype system locates the reader using a jig that significantly simplifies the alignment issues.

The user interface is a critical part of the design in such a system and may well determine whether the system is destined for "life in a cupboard" or is to become a valuable aid for the visually disabled person. The user interface is described in detail below.

The technological problems associated with the system are concerned chiefly with the derivation of robust image processing algorithms suitable for the application. The image processing algorithms are required to work with all commonly used display technologies and a wide range of display types. Other problems are concerned with the identification of a low-cost, low-power, small-sized image sensor having the appropriate level of spatial and contrast resolution for this application and the system design to support a good and useful user model.

From the above, one can see that the system needs to be highly integrated and to use some form of ASIC-based design. Yet there are two areas that require considerable levels of experimentation: in the image processing area, to prove display technology independence of both the image sensor and the display reader algorithms; and in investigating the usability of the system, both in terms of the user interface and in the packaging of the product. This need for experimentation is rather at odds with the high-cost development route associated with producing a highly integrated product. One could resolve the necessary experimental issues by using computer-based modeling techniques; where executable, prototype models of the system running on a computer system may be evaluated for their image processing capability and the suitability of the proposed user interface. Such a technique has been described (6),

albeit directed at another class of applications. These techniques are effective methods of assessing design issues before developing highly integrated (so-called deeply embedded) systems. However, in this project the development of a physical prototype was chosen.

The primary reason for choosing a physical prototype is so that the usability of the system can be evaluated by undertaking controlled experiments with visually disabled users. The authors felt that a physical prototype would be more appropriate for this exercise. A further, more pragmatic reason concerns the fact that this work was undertaken by a group of master of engineering degree students working on an assessed project of the type where rules demand that skills in both software and hardware engineering be demonstrated.

The project has been carried out with full consideration of a final usable product. Although size, power, and cost constraints have not been directly addressed in the construction of the prototype, they have been fully accounted for by the system design. The design is technology independent and may be reused in an embedded system implementation.

User Model

Although the fundamental requirement for this system is to enable a user to operate consumer electronic products, as has been stated above, it is critical that this is achieved in a way that results in the easy use of the product. The objective has been to make a reactive system in which the user operates a product in a manner that closely resembles that used by someone without a visual disability, and where the speech output from the system is given automatically when appropriate for a given context. This is illustrated below by considering user interaction with a typical VCR.

To assist in meeting the goals for the display reader, lessons can be learned from computer speech access systems for blind people. These "screen readers" enable blind people to operate standard applications running on IBM-compatible personal computers. There are many thousands of blind users of such systems around the world. Screen readers have developed from systems in which users could ask for information about areas of the computer screen (7), such as the current character, word, or line, into sophisticated systems in which configurations or profiles are constructed for given applications and that provide application-specific knowledge (8). Although users have the facilities to produce their own configurations, configurations for the major applications typically are made available by the companies who produce the screen readers. The importance of making such systems configurable and adaptable is detailed elsewhere (9), where the objective is stated as making the speech access system provide a seamless interface between application and user.

An important issue that should be noted when considering speech systems is that the access needs to be balanced with the quantity of information. "Minimum speech with maximum information" (10, p. 537) has been our philosophy in the design of the user interface for this system.

The approach for the display reader has been to construct a configuration for each display, which is set up so that, in most circumstances, the system automatically responds with an appropriate speech message as the user interacts with the consumer appliance. This is achieved by providing the facility to label fields as:

- *read field on change*, where the contents of the field, or one of a set of speech messages, will be spoken when a change occurs in that field
- *read field continuously*, where the field status will be repeatedly reported at a specified frequency
- *read once*, which is generated by a user request.

For each of these options the spoken text can be derived from the display field, a previously stored text string, or a combination of the two. For example, "The time is 10 p.m." where the information "10" is extracted from one field on the display, and the "p.m." is extracted from another field. In addition, an option is provided for a user to manually step through and examine different fields.

These capabilities can be illustrated by describing a user interaction with a typical VCR, where the user is to record a program that is about to start on the television set. Initially the user will attach the reader to the jig on the recorder and will insert a tape into the machine. The user now presses REWIND to ensure that the tape is at the start. The display changes and a "<<" symbol appears on the display that the display reader notices as a change and speaks the message "rewind." When the system stops, the display symbol "<<" is cleared and the system will announce "stop." At this point, the user may wish to reset the program counter and presses the COUNTER RESET button. This is not spoken and the user will need to manually operate the reader to step to the COUNTER field and read the contents. (The counter change is not automatically read on change as this would be very distracting in most circumstances). The user then presses RECORD and PLAY which again change the display and the system would respond by announcing "recording."

System Architecture

The Logical View

The system architecture is presented in **Figure 3** using structured analysis notation (11). This is a logical view of the architecture and is independent of implementation. The actual prototype system on which initial experimentation has been based is shown in **Figure 4**.

The way in which the system operates is briefly described with reference to **Figure 3**. The Image Sensor is placed in front of a display. In essence, the sensor is a video camera that continually passes video information to the system. The Capture Image and Image Store are essentially a hardware frame grabber, which digitizes the video signal and stores it as a pixel array in the Image Store memory. Although not implemented in the prototype system, where standard components have been used, there is a strong possibility that in the final system the Capture Image process may perform some elementary image processing operations (e.g., thresholding) "on the fly," as pixels arrive, and before they are placed in the Image Store. Images are captured when requested by the Capture Image flow; completion of image capture (i.e., an updated Image Store) is signaled by the Image Captured flow.

The image is then processed to determine the semantic information currently presented by the display. This is accomplished in two stages: Process Image, which operates on the raw pixel information to determine which areas of the display (cells) are illuminated; and Analyze Image, which operates on the list of cell states to determine the meaning of the information and whether this should be spoken. Both processes reference the Display Configuration Store. This store is divided into two parts: a static part, which contains configuration information for the particular display (e.g., the fields in the display, the locations of elements and of cells in the fields, the text associated with each field, etc.), and a dynamic part, which is a reflection of the current state of the user preferences and determines, for example, which set of fields should be examined, the format of the text to be spoken, etc. The image processing algorithm is described in detail later.

The System Control process determines when a frame should be captured and processed, and what, if any, speech should be passed to the user. A Capture Image message is sent either in response to some user command that explicitly requests that a display field or fields, be read, or is generated based upon the current operating mode of the system, as recorded in the Display Configuration Store. For instance, when the system is configured to "read field on change" (see the section above on User Interface), frames must be periodically grabbed at a rate exceeding one frame per second. The User Interface carries out text-to-speech synthesis on strings of text and processes commands received from the user via the keypad.

The image processing functional units, Process Image and Analyze Image, rely upon having access to the unique characteristics of the display held in the Display Configuration Store; this is the static part of the store mentioned above. This information is used by Process Image to locate

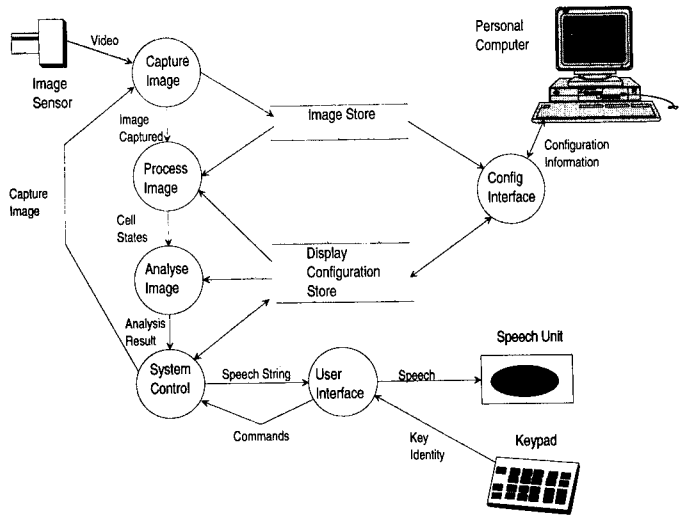

Figure 3.
The system architecture.

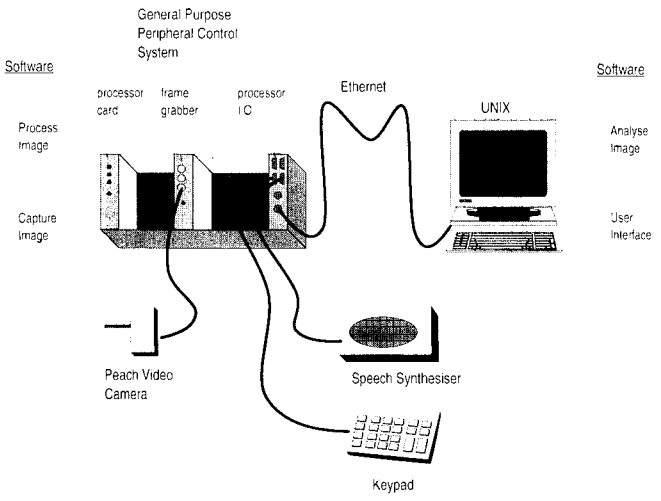

Figure 4.
The display reader prototype system.

fields within the Image Store and to determine the format of those fields. Analyze Image uses this information to determine the text strings to be spoken. Thus, for each display to be read there is a configuration entry in the Display Configuration Store to supply this information to these processes. This information is required each time the display reader reads a particular display and is therefore held in nonvolatile memory. There must be some way of capturing the configuration information for each display; which, it is anticipated, will have been produced by a sighted operator. This is supplied through the Config Interface, which is a process that needs access to both the Image Store and the nonvolatile part of the Display Configuration Store. The issues relating to system configuration and the role of the sighted operator are discussed later.

The Prototype System

The prototype system, on which experimental work has been carried out, is briefly described in this section, and is illustrated in **Figure 4**. The display reader system prototype has been developed using a General Purpose Peripheral Control System (GPPCS) interfaced to a UNIX workstation. This environment (12) allows an application programmer to develop prototypes of embedded systems using the features of the UNIX operating system. In general, applications programs are written for, and reside in, the UNIX workstation. In this prototype system the Process Image, Analyze Image, and User Interface were developed in the UNIX system and interfaced to the relevant input/output (I/O) devices through peripheral dialogue library (PDL) functions. These functions present a high-level interface to the application software. For instance, the PDL function for the speech synthesizer will accept strings of text as parameters. The PDL functions manage communication with the GPPCS and call the relevant device driver library (DDL) in the GPPCS; at this level of abstraction the speech synthesizer is controlled by an RS232 driver procedure. The scheduling of tasks within the GPPCS is controlled by a small kernel that runs on a 68030-based processor card and controls the I/O devices over a VME bus.

In most circumstances, the communication overhead between the GPPCS and the UNIX workstation is insignificant. However, in the display reader prototype the Process Image process has relatively low CPU overhead but operates on the whole captured image. In this case, the overhead associated with shipping the entire image to the UNIX workstation and processing it there was about four times that of running the Process Image procedure within the GPPCS and transferring the cell states to the Analyze Image process running in the workstation.

The prototype system acts as a sound platform for the development of the software for the system, which could be ported with little modification into an embedded system product. The prototype does not reflect the hardware architecture of a portable system. However, the prototype system was used to evaluate the most critical hardware component, namely, the image sensor.

The image sensor must be of low cost and small size; have minimal power requirements and sufficient spatial and contrast resolution for the application; and provide acceptable quality images over a wide range of display types. The first three of these requirements are largely satisfied by the "Peach camera" manufactured by VLSI Vision (Edinburgh, UK). This is a 312×287 pixel image sensor array with on-chip circuitry to deliver a full-format composite video output signal. The ASIC Image sensor (13) may be purchased as a die, mounted on a printed circuit board (PCB) or packaged as a miniature camera, complete with wide angle lens. The latter option was used in this project; the entire video camera measures $3.3 \times 3.5 \times 2.7$ cm. The camera is of relatively low cost (around £50, approximately, U.S. $75, per unit in volume) and has good power characteristics, dissipating less than 150 mW. The camera, therefore, appears well suited to this application. Evaluation of the camera was carried out in the prototype environment using a number of display types; the results of this evaluation are summarized in the results section below.

METHOD

Image Processing

A robust image processing algorithm is required to read the wide range of display types available in the domestic market. As the display reader is required to be held in a jig immediately in front of the display, ambient lighting conditions can be controlled. For emitter displays (LEDs and GPDs), the emitted light is uniform from all areas of the display and these displays have sufficient contrast between lit and nonlit areas for simple adaptive thresholding techniques to be used. Nonemitter displays (i.e., LCDs) are available in two types: backlit, where the emitted light from the display gives sufficient illumination for treatment in the same way as emitter displays; and nonbacklit, where a light source must be provided by the display reader unit to illuminate the display. In the latter case, careful design

of the unit is needed to ensure reasonably uniform lighting of the complete display and the elimination of reflection from the surface of the display. In the prototype system jigs have not been used, the camera has been held in front of the display and ambient light has been used to illuminate nonemitter displays. Further work is necessary in jig design to develop satisfactory lighting conditions.

Standard optical character recognition (OCR) techniques are not applicable for reading product displays. In order to determine the meaning of symbols (static elements) within the display, the algorithm must refer to a set of configuration details for the display. For instance, a single dot on a display, which in another field presents time, may indicate that the displayed time is p.m. (or conversely a.m., there is no standard convention). An entry must exist in the Configuration Display Store for this information to be correctly interpreted. Thus, the characteristics of displays force the system to have a configuration entry for each display. However, once this principle is accepted, it actually makes the design and implementation of the image processing method rather more simple than if conventional OCR techniques were used. This is because the configuration entry can indicate to the Process Image function the set of cells, elements, and fields in the display, their configuration, and their precise position in the display. In the limit, this means that, if the camera is aligned precisely in the same way as when the configuration details were obtained, the image processing task becomes a reasonably simple task of checking various areas of the screen to see whether they are emitting light.

The prototype system relies on precise alignment (i.e., for the system to work the camera must be in the same position relative to the display as when the configuration details were captured). In the final system such alignment would be difficult to attain. Each domestic appliance in the user's house would need a jig to hold the display reader in front of the display. Locating the camera accurately in the jig each time is not a difficult problem to solve; however, it is likely that such a system may be difficult to use and may include potentially costly financial overheads both for manufacture and installation. However, the idea of a jig is appealing because it gives the user hands-free operation of the display reader; ensures that the display is entirely within its field of view; and holds the image sensor at a fixed distance from the display, thus maintaining a fixed character size. Therefore, it is assumed that a jig will provide some form of coarse, first-order alignment of the display within the field of view. For instance, the jig may be a number of Velcro™ strips around the product's display that interlock with similar strips on the display reader. A software process will align the current display image with that captured in the configuration store. Given that the precise nature of layout of the display is known, this is not a difficult task. In fact, a simple alignment program was developed for the prototype system; this worked well in principle but was defeated to some extent by the "fishbowling" introduced into the image by the camera (see the results section below).

The image processing method is now briefly described; it assumes that accurate alignment is maintained.

The configuration entry for the display identifies the fields on the display, the type of elements, which make up the (fields) and the precise location of the cells, which form the elements. For the low-level image processing algorithm the primary unit of concern is the element. These were defined earlier and can be one of four formats: display symbol, 7-segment character, 16-segment character, or dot matrix character. By knowing the size and format of the element, the exact position, shape, and size of the cells that form the elements can be determined. Thus for a 7-segment display, each of the seven cells that make up the display can be tested to determine if they are illuminated. Illumination of a cell is determined by thresholding the image, counting the number of illuminated pixels within the cell area, and comparing the value to some defined threshold. This process is illustrated in **Figure 5**.

The list of cell states for each element is returned, together with the field and element identifiers, for subsequent processing. From the list of cell states, characters can be identified. This is shown for a 7-segment display in **Figure 6**. Similar lists can be developed for other field types. Note that the term character is used rather loosely and does apply to elements formed from display symbols.

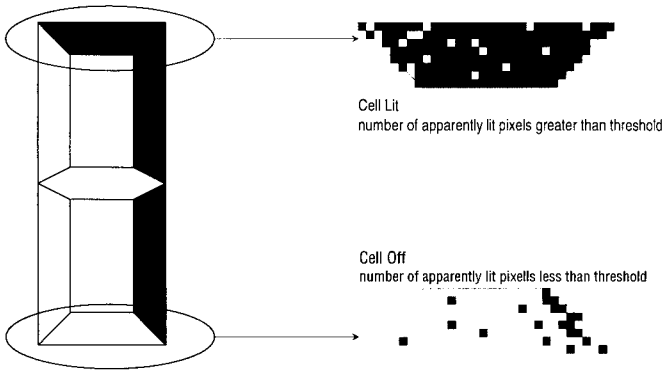

Figure 5.
Determination of cell illumination.

Note also that this algorithm works for dot matrix characters where the position of the characters is fixed on the display; however, where characters are of variable size, or can start at any point in the display, other algorithms must be used in conjunction with the one described here.

For each field, the character displayed by each element can be determined from the cell list. Dependent on the type of information held in the field the spoken phrase can now be constructed. A simple example of an alarm clock is shown in **Figure 7**. The alarm clock display consists of three fields: 1) a time field made up of four 7-segment elements; 2) an a.m./p.m. field which is a single static element (this is lit to indicate "p.m."); and, 3) an alarm set field which is a single static element (this is lit to indicate that the alarm is set). To speak the time, the display reader must chain together the phrases associated with the time and the a.m./p.m. fields. The alarm set field is spoken independently and is not considered further here.

The phrase for the time field is held as: "The time is ~." The character ~ indicates that the contents of the field should be spoken at this point. Thus, the field contents can be inserted into any position in an arbitrary text string. The field is marked in the configuration store as being time format; thus, the set of characters 1 2 3 4, is spoken as "twelve thirty-four," rather than "one thousand two hundred and thirty-four," which would be the case if the field was marked as a value field. Field 2 (a.m./p.m.) is marked as an alternative field, where two alternative phrases are stored, the phrase used being dependent on the state of the field: "pee em./aye em."

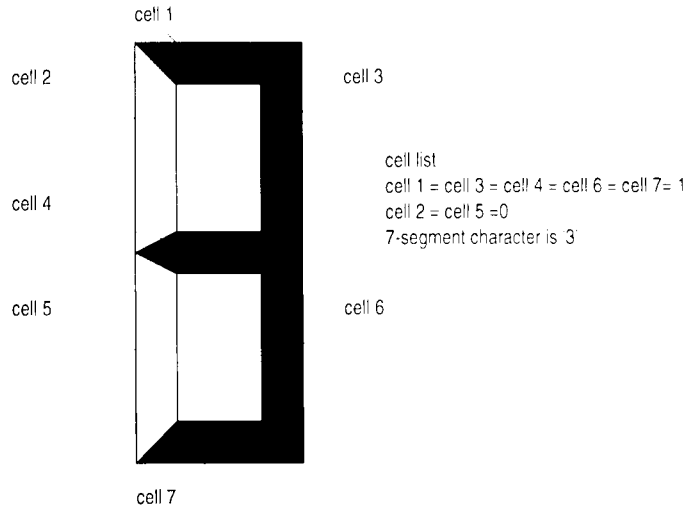

Figure 6.
Determination of cell states for a 7-segment display character.

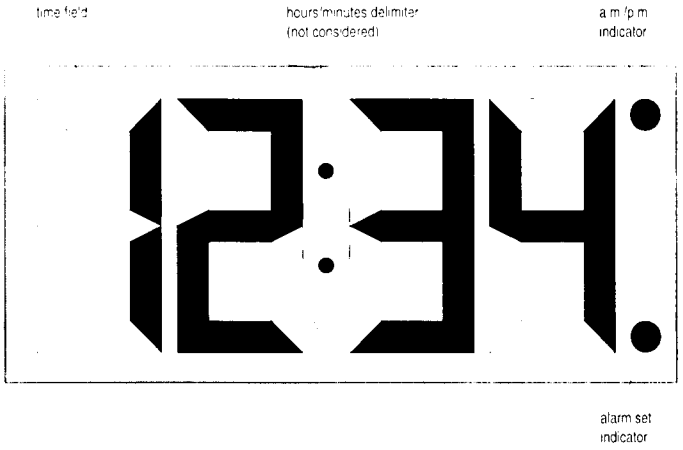

Figure 7.
An alarm clock display.

Dependent on the current operating mode of the system (see User Interface section above), which is held in the Display Configuration Store, and on the values obtained from the display, the text strings may or may not be passed to the speech synthesizer unit. Considering the alarm clock example, if the mode was set to read field on change, the reader system samples the display about once per second and compares the value of the time field with the value captured previously. Text is spoken when a change is detected; in this example text would be spoken every minute. If the user requested that a field be spoken, the field would be returned and spoken immediately.

RESULTS

Prototype Evaluation

Display Quality

The quality of the captured image has been evaluated by capturing images from the displays of a number of domestic products, which cover the range of available display technology (i.e., LEDs, GPDs, and LCDs) both backlit and lit by ambient light. The quality has been assessed by visual inspection following the thresholding of the image; in all cases lit cells were clearly distinguished.

System Evaluation

The prototype system has been evaluated by operating the system with three domestic products. Again, a sample representative of the range of image technologies was chosen; namely LED, GPD, and ambient lit LCD. The system has also been evaluated by running simulations of

product displays on portable computer systems. This allows the emulation of complex displays, yet gives total control and repeatability to each of the experiments. Most of the work to date has been carried out using an emulation of a video recorder, which has representatives of all the elements found in the complete set of domestic products. This emulator has been run on two PCs; one with a GPD and the other with backlit LCD. This testing has verified that the image processing algorithm is robust and virtually error free over a reasonable range of display technologies. One small error has manifested itself, however. When reading displays in *read field on change* mode the display image is sampled once every second, for 7-segment displays the character "0" is read in error, on average once every 20 minutes, as the character "8." This manifests itself by the system indicating that the channel of the emulated video recorder display has changed from "channel 2" to "channel 82" and on the next read back to "channel 2." This is equivalent to an error rate of approximately 0.08 percent. Only 7-segment zeros seem to suffer from this defect.

Image Alignment

As noted above, the image processing algorithm requires that the camera and display be in the same relative alignment as when the system was configured. System configuration takes a significant period of time, around 20 minutes for a video recorder display. Any movement of the camera relative to the display, even if this movement is comparatively small, results in the system having to be reconfigured.

As noted earlier, such precise alignment will be difficult, if not impossible, to maintain for a practical system. Therefore, work will be undertaken to permit the system to align a captured image with the information held in the configuration store.

Fishbowling of the Image

The image alignment problem is compounded by the fact that the image passed by the Peach camera to the system is fishbowled (i.e., straight segments at the edge of the image appear curved). This is because the Peach camera has a wide-angle lens, and in this application the camera is in rather close proximity to the display. This problem could be overcome by either obtaining a miniature close-up lens for the Peach camera, or by applying some sort of a corrective transformation to the captured image.

System Configuration Policy

As has been discussed above, the system will need a specific configuration for each appliance with a display. It is envisaged that, although facilities will exist for a sighted user to produce configurations for a given individual, the configurations generally will be produced and distributed by a central agency, probably the supplier of the display reader. Although this will involve a considerable effort initially, it is expected that subsequent efforts will only involve keeping up to date with the latest product developments. There are issues relating to upgrading individuals with configurations for the latest devices; however, this is well within current capabilities for electronic data transfer and is not dealt with here.

DISCUSSION

The ultimate aim of this work is to produce a hand-held product, at a reasonable cost, that is suitable for use in the home. To this end four areas of further work are under consideration:

1. **Testing with a Wider Range of Products**. As outlined in the section concerned with prototype evaluation above, a limited yet representative subset of displays has been used to evaluate the system. More extensive testing is required and a systematic evaluation of the prototype system will be carried out with a larger set of displays.
2. **The Evaluation of the User Interface**. It is intended that the user interface will be evaluated by carrying out a series of experiments with visually disabled users of the system. The purpose of the work is twofold: to ensure that a suitable user interface is developed and to develop a more general level of understanding as to how visually disabled people can effectively operate with modern domestic appliances.
3. **Image Alignment**. Further work is necessary to address the image alignment problems and fishbowling discussed above. It is anticipated that this work will include experiments conducted in a similar manner to those in the previous section, with visually disabled users of the system.
4. **Development of the Target System**. Satisfactory completion of this work will result in a prototype of a highly integrated, hand-held product that can be used in the home. This will entail developing an ASIC-based system. This work will progress in par-

allel with other, related work undertaken by the systems engineering group at UMIST as part of the ESPRIT OMI/DE[1] project. It will result in the production of a deeply embedded processor-based system for a range of image processing applications, of which the system described here is one. The system will be based around an ARM[2] processor macrocell; this processor has high computational power and therefore is capable of meeting the real-time requirements of the display reader system and has low power requirements.

CONCLUSIONS

A proof of concept demonstrator has been constructed and sufficient evaluation carried out to generate confidence that this approach to the problem of visually handicapped people accessing visual displays is valid. Despite its limitations, the prototype system can and should be used to evaluate the real needs of visually disabled users. As for subsequent development at the target price of $100 suggested by the World Blind Union Research Committee, this seems unlikely in the short term. However, improvements in silicon processing technology will allow a system to be integrated onto a single silicon chip in the future and would probably enable the system to be supplied at its target price in from 4 to 5 years from now.

ACKNOWLEDGMENTS

This project was carried out by a group of five students as part of their Master of Engineering degrees in microelectronic systems engineering or in software engineering, at UMIST in the academic year 1992/93. Joe Carter, David Gent, Justin Peters, Chris Ross, and David Whipp developed the system from initial specification supplied by the Technology for Disabled People Unit (TDPU). In transforming this specification into the prototype system, these students have shown tremendous enthusiasm and innovative ability, and have put in a great deal of hard work. The authors thank them for their assistance. We would also like to acknowledge the major contribution to the work reported here, and to the TDPU in general, of our colleague Professor Derrick Morris. The TDPU would like to acknowledge research grant funding from the Guide Dogs for the Blind Association, which was used to fund this work.

[1]The European Community has a funding program for precompetitive research in the information technology industries to improve Europe's competitiveness. This is known as ESPRIT, part of which is concerned with the microprocessor/microelectronics sector and known as the Open Microprocessor Initiative (OMI). The Deeply Embedded(DE) project is investigating highly integrated systems that result in complete software/hardware systems on a single silicon chip.

[2]This refers to a family of high performance, low power consumption 32-bit RICS processors developed in the United Kingdom by ARM Ltd. The technology is licensed to a number of silicon vendors worldwide and the processors have been used in a large number of high-performance, battery-powered devices.

REFERENCES

1. Kay L. Electronic aids for blind persons: an interdisciplinary subject. IEEE Proceedings 1984:131(Part A):559–76.
2. Meyers A, Schreier E. An evaluation of speech access programs. J Visual Impairm Blindn 1990:84:26–38.
3. Schwerdtfeger RS. Making the GUI talk. BYTE 1991:December:118–28.
4. Boyd IH, Boyd WL, Vanderheiden GC. The graphical user interface: crisis, danger, and opportunity. J Visual Impairm Blindn 1990:84:496–502.
5. Gill JM, ed. Priorities for technical research and development for visually disabled persons. World Blind Union Research Committee, 1992.
6. Morris D, Evans DG. Modelling parallel and distributed systems. Parallel Comput 1992:18:793–806.
7. Blenkhorn P, Caulderwood D. Access to personal computers using speech synthesis: a review of the past decade. The New Beacon—Blind Welfare 1992:LXXVI(898):185–8.
8. Ronsse C. The IBM workstation for the blind. In: Francois G, Engelen J, eds. Proceedings of the 6th International Workshop on Computer Applications for the Visually Handicapped, Leuven, Belgium, 1990.
9. Talbot M, Blenkhorn P. Building context-sensitivity into special access products. In: Murphy H, ed. Proceedings of the Seventh Annual International Conference on Technology and Persons with Disabilities, 1992 March, Los Angeles, CA.
10. Vincent T, Turnbull SD. Word processing for blind people. Microprocessors Microsystems 1984:8:535–8.
11. Yourdon E. Modern structured analysis. Englewood Cliffs, NJ: Prentice-Hall, 1989.
12. Carter JA, Gent DP, Peters JL, Ross CC, Whipp DP. Display reader for the Blind. Microelectronic System Engineering and Software Engineering MEng., Project Report. Manchester, UK: University of Manchester Institute of Science and Technology, 1993.
13. Renshaw D, Denyer PB, Wang G, Lu M. ASIC image sensors. In: Proceedings of the IEEE International Symposium on Circuits and Systems, 1990.

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CLINICAL REPORT

Clinical and Laboratory Study of Amputation Surgery and Rehabilitation

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INTRODUCTION

Prosthetics Research Study (PRS) continues its ongoing programs of clinical and laboratory research into amputation surgery; immediate postsurgical amputee management; lower limb prosthetic development, including the automated fabrication of prostheses and other mobility aids; a basic and clinical study of wound healing as it relates to amputation surgery and postsurgical management; engineering investigation into the mechanical properties of soft tissues and the response of living soft tissues to application of force; and technology transfer of automated fabrication methods into the clinical services of the Department of Veterans Affairs medical care system.

PROGRESS AND RESULTS

Prosthesis Force Transducer

Preliminary design and engineering analyses of the prosthesis force transducer have been completed. The design incorporates piezoelectric quartz crystals as force sensing elements to obtain force information at three discrete locations within the transducer. Given these data (9 forces), the complete set of forces and moments (Fx, Fy, Fz, Mx, My, Mz) at the socket due to ground reaction forces can be obtained. The device as designed will be less than 1 cm thick, 7 cm in diameter, and will weigh approximately 800 g. To date, the analog electronics have been purchased as well as the crystals and other raw materials.

Diabetic Footwear

In conjunction with the PRS diabetic footwear study, we have undertaken a study of three-dimensional (3-D) foot morphology. We have obtained digital images of over 100 feet using a Cyberware™ laser scanner, and over 50 pairs of images from an Amfit™ contact scanner. The laser-scanned images provide full 3-D shape information over the entire foot (dorsal and plantar aspects), while the contact scans provide only plantar surface information. Custom software has been written and the DVA/ShapeMaker™ software has been modified to facilitate quantitative comparison of these scans. To date only preliminary analysis of the data has been undertaken, but shows great promise as a tool for understanding foot morphology, as well as designing and fabricating custom insoles and footwear.

Gait Activity Monitor

We have developed a compact, self-contained gait activity monitor (GAM) which records the number of steps taken by a patient over a 2-week period. The GAM does not require patient intervention and has a sealed, waterproof case which prevents tampering. It provides the clinician with an objective, reliable measure of functional outcome for evaluation of new prosthetic devices and medical treatments. We have built four prototype units and collected data from three subjects for up to one week. We are currently refining the mechanical sensor and designing software for data analysis.

For further information, contact Ernest M. Burgess, MD, Prosthetics Research Study, 720 Broadway, Seattle, WA 98122.

The research and development in these studies was sponsored by the Department of Veterans Affairs Rehabilitation Research and Development Service, Washington, DC.

Mechanical Properties of Skin

To study the response of skin to mechanical stress, we designed and built an automated mechanical stimulator capable of applying precise, repetitive forces in both the normal and shear directions. The stimulator was applied *in vivo* to the skin of pigs with gradually increasing amplitude over a specified number of days. At the end of the trial period, tissue samples were taken and studied using standard histological procedures. Preliminary experiments suggest there is a change in the structure of collagen fibers in response to the mechanical stress.

Residual Limb Volume Changes

An optical silhouette scanner has been designed and constructed for measuring both diurnal and long-term volume changes in a residual limb. This device utilizes a rotating charge coupled device (CCD) camera and light source to obtain a series of 2-D silhouettes around the residual limb. The silhouettes are then reconstructed into a 3-D computer image of the residual limb from which volume measurements can be made. The device is currently under preliminary testing.

Studies were completed where it was found that gamma irradiation of blood transfusions inhibited their ability to sensitize to minor (non-major histocompatibility) transplantation antigens as part of a project to modify blood so as to prevent sensitization yet maintain the ability to induce tolerance for foreign transplant antigens.

AFMA

The Automated Fabrication of Mobility Aids (AFMA) system makes it possible to produce a limb at reduced price in a shortened time. Rectification techniques permit a consistent socket fit. The data are easily filed with reproducibility of the limb and any needed adjustments easily, quickly, and inexpensively made. These techniques now include a PRS above-knee (AK) socket design unique in its characteristics in that it incorporates the advantages of both the classical quadrilateral socket, the narrow ML (Long) socket and the advantages of a flexible/external frame socket. This technique together with the development of the VA/DAV/PRS Knee has allowed us to complete the lower limb prosthetic system as planned.

Training courses were developed and begun for the technology transfer of AFMA into the VA Medical Centers. Three centers were established on the west coast, two remote sites where AFMA limbs are now designed and fit and one central fabrication site where they fabricate the AFMA sockets.

BIBLIOGRAPHY

Burgess EM, Rappoport A. Physical fitness: A guide for individuals with lower limb loss. Baltimore, MD: Rehabilitation Research and Development Service, Department of Veterans Affairs, 1993.

Houston VL, Burgess EM, Childress DS, et al. Automated fabrication of mobility aids (AFMA): below-knee CASD/CAM testing and evaluation program results. J Rehabil Res Dev 1992:29(4):78–124.

Kuroki H, Bean MA, Ikuta Y, Akiyama M, Burgess EM. Effect of FK506 and donor-specific blood transfusion on the rat composite tissue limb allograft and the mechanism of long-term graft survival. Transplant Proc 1993:25(1):658–61.

Kuroki H, Bean MA, Ikuta Y, Burgess EM. Effect of FK506 and donor specific transfusion (DST) on skin and composite tissue allograft (CTA) survival in the rat. In: Transactions of the 38th Annual Meeting of the Orthopaedic Research Society, 1992, Washington DC, 1992:17(1):176.

Mathews D, Burgess E, Boone D. The all-terrain foot. J Prosthet Orthot 1993:5(1):29–44.

Sanders JE, Daly CH, Burgess EM. Clinical measurement of normal and shear stresses on a trans-tibial stump: characteristics of waveform shapes during walking. Prosthet Orthot Int 1993:17(1):38–48.

Sanders JE, Daly CH, Burgess EM. Interface shear stresses during ambulation with a below-knee prosthetic limb. J Rehabil Res Dev 1992:29(4):1–8.

Smith DG, Burgess EM, Zettl JH. Special considerations: fitting and training the bilateral lower limb amputee. In: Bowker JH, Michael JW eds. Atlas of limb prosthetics: Surgical, prosthetic and rehabilitation principles. 2nd ed. St. Louis, MO: Mosby Yearbook, 1992:599–622.

Smith DG, Stuck RM, Ketner L, Sage RM, Pinzur MS. Partial calcaneectomy for the treatment of large ulcerations of the heel, and calcaneal osteomyelitis: an amputation of the back of the foot. J Bone Joint Surg 1992:74A(4):571–6.

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CLINICAL REPORT

Who Is At Risk of Limb Loss And What To Do About It?

An Invited Presentation at the VA Rehabilitation Conference, sponsored by Prosthetics and Sensory Aids Services, VACO, Frederick Downs, Director, April 28, 1994, Las Vegas, NV

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Abstract—Lower limb amputations were performed on over 105,000 individuals in United States short-stay hospitals between 1989 and 1992. Additional amputations were performed in VA, military, Indian Health, and charitable orthopaedic hospitals. Half of all lower extremity amputations occurred in individuals with diabetes. When the causal chain leading to diabetic amputations was examined in 80 consecutive patients at the VA Medical Center, Seattle, WA, 23 unique pathways were identified. Multiple pathway components were identified for 96% of patients, while in 4% a single ischemic pathway was sufficient in itself to require amputation.

The majority of the scenarios leading to amputation began when patients with absent peripheral sensation sustained a pivotal event that initiated the causal chain to amputation. In nearly half the patients, this event was footwear-related. The pivotal event was followed by ulceration and faulty wound healing in 73% of patients.

Each year thousands of individuals with diabetes undergo amputation in VA facilities, resulting in substantial cost to the Department of Veterans Affairs and to themselves. If the VA is to address the prevention or delay of limb loss, the causal pathway information indicates that attention to the footwear of diabetic patients is necessary.

Key words: *amputation, causal pathway, diabetes, footwear, ulcer prevention.*

INTRODUCTION

On average, 109,411 individuals underwent amputation in United States (US) short-stay hospitals between 1989 and 1992. Approximately 96 percent of these individuals amputations were lower limb and 4 percent were upper limb. Thousands of additional amputations were performed at military, Department of Veterans Affairs (VA), Indian Health, and charitable orthopaedic hospitals. Indications for amputation corresponded with the age of the individual, such that younger individuals experienced more congenital-, malignancy-, and trauma-related amputations, compared to older individuals whose amputation codes reflected multiple pathophysiologic mechanisms (e.g., diabetes, ischemia, and infection). During 1989–1992, 51 percent of the lower limb amputations in US short-stay hospitals occurred in individuals with diagnosed diabetes, yet persons carrying the diagnosis of "diabetics" represent only 2.5 percent of the US population (1). Be-

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tween 1980 and 1990, a 29 percent increase in the age-adjusted diabetic amputation rates was noted. These amputation rates were uniformly higher in males than in females, in blacks than in whites, and increased with advancing age. The length of hospitalization for amputation fell from 35.8 days to 20.6 days during this interval (2).

Causal Pathways and Risk Factors For Diabetic Amputation

To determine why individuals with diabetes experience excessive lower limb loss, important pathophysiologic factors and other events leading to amputation need to be identified. The causal pathway model described by Rothman has been used for this purpose (3,4). The two major components of the model, "sufficient" and "component" causes, are illustrated in **Figure 1.** A sufficient cause is a set of minimal conditions and events that inevitably produce the outcome with no superfluous conditions or events. In disease etiology, when the components assembled form a sufficient cause, the onset of disease occurs. A component cause is not sufficient in itself to cause disease but is one of several components or pieces that must be assembled for the disease to occur. The Rothman model allows assimilation of empiric information from representative cases and considers different contributions along a disease course from the time of the initiating event to the addition of the final component cause. The critical public health message imparted by this model is that preventing the occurrence of a component cause stops its entry into the pathway and thus renders other components unable to produce the outcome (3,4).

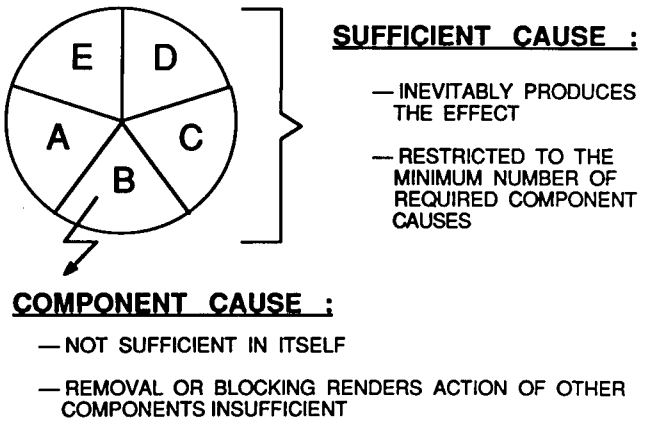

Figure 1.
Diagram of sufficient and component causes. A–E represent causes that are not sufficient in themselves but that are required components of a sufficient cause that inevitably produces effect. Adapted from Rothman (3). (Reprinted, with permission, from Diabetes Care, Vol. 13, No. 5, May 1990.)

METHOD

The causal pathway methodology was applied to diabetic amputations. The components included four major pathophysiologic mechanisms: neuropathy, ischemia, infection, and wound-healing failure; two common soft tissue complications, cutaneous ulceration and gangrene; and an environmental initiating event, (e.g., minor trauma, trauma from repetitive mechanical pressure or footwear). These components were selected based on historical and quantitative data and the diabetic amputation literature.

For this study, extensive health, physical and laboratory measures, medical history, and lesion photographs were uniformly collected and compiled into a standard patient work sheet. These measures were applied to the 80 consecutive diabetic male veterans, ages 30–85, who required their first amputation of an affected lower extremity for nontraumatic indications between 1984 and 1987 (5). Similar data were collected from a concurrent control population of 236 diabetic male veterans who underwent elective surgical procedures, not related to diabetes, performed by surgeons from eight subspecialty services. Cases and controls were jointly analyzed to control for age, race, diabetes type, duration, severity, and socioeconomic status; the results of this case-control study are described elsewhere (6).

The three causal pathway investigators independently assigned component and sufficient causes for the amputation based on guidelines that had been agreed upon in advance.

RESULTS

Final consensus was achieved among investigators using a modified Delphi process (7). The analysis identified 23 unique amputation causal pathways among the 80 cases. In 77 amputations, multiple components were reflected in the causal pathway; while in one pathway, which accounted for three cases, an ischemic pathway was sufficient in itself to require amputation. In eight pathways, which accounted for 73 percent of the amputations, the combination of minor trauma, ulceration, and faulty wound healing was present. **Figure 2** diagrams a causal pathway that includes baseline

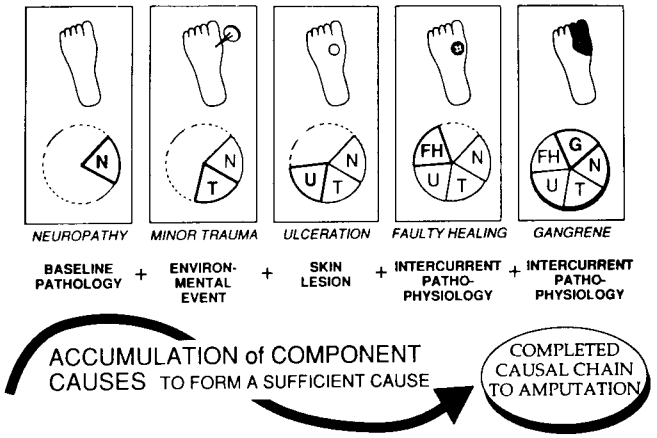

Figure 2.
Representation of causal pathway to individual amputation, which includes essential contributions from underlying diabetes-related pathophysiology (neuropathy), initiating environmental event (minor trauma), formation of foot lesion, and subsequent healing complications. Eventual occurrence of gangrene is terminal event of this causal chain, which requires participation of all preceding components before becoming sufficient to cause amputation. Theoretically, amputation could have been avoided by elimination of any one component cause before convergence of causal chain. (Reprinted, with permission, from Diabetes Care, Vol. 13, No. 5, May 1990.)

neuropathy, a minor traumatic pivotal event, ulceration, and faulty healing; the final component cause, which completed the causal chain, was gangrene.

The component causes present in the causal pathways leading to amputation included ischemia, 46 percent; infection, 59 percent; neuropathy, 61 percent; faulty wound healing, 81 percent; ulceration, 84 percent; and gangrene, 55 percent. The major risk factors identified for diabetic lower limb amputation from the case control-study were lack of vibratory perception on either lower extremity, low levels of cutaneous circulation, a history of peripheral vascular disease, low levels of high density lipoprotein subfraction 3, and lack of outpatient diabetes education (6). The population at attributable risk related to absence of vibratory perception was 72 percent and suggests the percent of disease incidence in the population of diabetic individuals due to this risk factor. A pivotal antecedent event or component cause that triggered the series of events ultimately leading to amputation was identified by 86 percent of the cases. For nearly half, the pivotal event either was shoe-related (42 percent) or might have been averted had the patient been wearing appropriate footwear (8 percent).

DISCUSSION

The Frequency of Amputation Risk Factors and the Magnitude of the VA Amputation Problem

In a study of risk factors for diabetic foot ulcers in the General Internal Medicine Clinic at the Seattle VA Medical Center (VAMC), the reported annual foot ulcer incidence was 5.7 percent[1]. **Table 1** shows interim self-reported findings on the prevalence of lower limb symptoms and procedures from a multi-site VA Cooperative Health Services Study of 732 diabetic individuals (8). Numbness of the feet or legs was reported by 60 percent, while over 40 percent reported no pain sensation in their feet. In **Table 2,** these patients identified their educational needs relevant to prevention of limb loss by indicating they had been taught nothing, a little bit, or that they wanted to know more about a topic (8).

Table 1.
Self-reported frequency of lower limb symptoms and procedures.

Symptom	Frequency (%)
Numbness of the feet or legs (sometimes, often, all the time)	59.7
No pain sensation in feet (sometimes, often, all the time)	41.5
Thigh/calf pain when walking (sometimes, often, all the time)	62.3
History of sores on feet/legs	28.0
Prior foot/leg blood vessel surgery	8.4
Prior amputation of toe, foot, or leg	7.0

Subjects: 732 diabetic patients attending three VA General Internal Medicine clinics.
Source: Cooperative Study in Health Services #7, Hines Center for Cooperative Studies in Health Services, Edward Hines Jr. VA Hospital, Hines, IL.

There are several hundred thousand diabetic veterans now receiving VA care. The lack of pain sensation results in many of these patients having no indication of lower limb pain, pressure, or trauma. Thus, these individuals are more likely to develop skin ulcerations in traditional shoes due to unperceived repetitive mechanical and shear stress. This lack of sensation, coupled with other complications, such as vision loss and limited dexterity for foot exams,

[1]E. Boyko: Personal communication, 1993.

may limit many patients in their ability to perform adequate self-care.

Table 2.
Self-reported frequency of educational needs.*

Educational Need	Frequency (%)
Checking feet regularly	27.5
Selecting proper shoes	38.6
When to call a provider	41.5
Whom to call for an urgent problem	39.6

Subjects: 732 diabetic patients attending three VA General Internal Medicine clinics.
Source: Cooperative Study in Health Services #7, Hines Center for Cooperative Studies in Health Services, Edward Hines Jr. VA Hospital, Hines, IL.
*Educational need is defined as knowing nothing, a little bit, or would like to know more.

The frequency of nontraumatic lower limb amputations in the VA was assessed for fiscal years (FY) 1986–1990[2]. **Table 3** shows a decline in diabetic and nondiabetic amputations during this interval. These data were used to estimate the economic impact of diabetic foot ulcers and amputations in VA. Direct and indirect costs for outpatient and inpatient ulcer care reflect current costs at the Seattle VAMC. Private sector diagnostic related group (DRG) reimbursement figures were used to estimate amputation and revision cost since VA is a "non-priced" setting (9). The frequency of serious foot pathology progressing to amputation was estimated at 15 percent. To extrapolate backward, the reciprocal of 15 percent, or 6.667, was multiplied by the average number of diabetic amputations (9,000) to yield an estimated 60,000 annual "pathologic lesions." Of the serious foot pathology, which includes foot ulcers, 20 percent were estimated to require inpatient care and the remainder required outpatient care averaging 4.5 visits per veteran. Although private length-of-stay figures were used for amputations and revisions, it is noteworthy that in statewide California data, the average amputation length-of-stay was 21 days in 1984 and 18 days in 1987 (10). This contrasts with the VA FY 1984 figure that 64 percent of amputation hospital stays exceeded 30 days and 6 percent surpassed 180 days[2]. Cost assumptions for **Table 3** would be modified by changes in amputation frequency, length-of-stay, and the addition of other diagnostic and surgical procedures (e.g., revascularization). Unique VA costs such as changes in pension, relocation to VA-supported extended care facilities and extended rehabilitation, equipment, and services are not included. Based on the above assumptions, an estimate of the annual VA diabetic foot ulcer and amputation costs is shown in **Table 4** and indicates that annual costs now exceed one-third billion dollars.

[2]Personal communication: J. Taylor, Biometrics Division, Management Sciences Services, Office of the Assistant Secretary for Plans and Policy, Department of Veterans Affairs, Washington, DC 1992.

Table 3.
Nontraumatic lower limb amputations,* in Department of Veterans Affairs Hospitals, for fiscal years 1986–1990.

Fiscal Year	Diabetes, ICD 84.10–84.18	Diabetes, ICD 84.3 Revision	No Diabetes, ICD 84.10–84.18	No Diabetes, ICD 84.3 Revision
1986	9,944	1,384	8,516	1,322
1987	8,942	1,367	8,407	1,354
1988	8,910	1,323	8,296	1,288
1989	8,638	1,229	7,997	1,190
1990	8,551	1,159	7,899	1,128
TOTAL	44,985	6,462	41,115	6,282
Average	8,997	1,292	8,223	1,256

* = Number of procedures, not individuals.
ICD = International Classification of Disease Codes.
Source: Compiled from VA Patient Treatment File.

Table 4.
Estimated annual costs of diabetic foot ulcers and amputations in the Department of Veterans Affairs.*

Estimated Number	Procedure	DRG	Estimated Cost ($)
60,000	Foot ulcers and other pathology		
12,000	20% inpatient care,* 8 days @ 1,078/day**	271	103,488,000
48,000	80% outpatient care,* 4.5 visits @ 95/visit		20,520,000
10,300	Amputations and revisions***		
3,600	Toe/Foot @ 15,124	114	54,446,400
5,400	BK, AK @ 33,444	113	180,597,600
1,300	Amputation revisions @ 18,495	213	24,043,500
TOTAL			383,095,500

*FY 1994 direct and indirect cost estimates from the Seattle VA cost accounting system for ulcers.
**This estimate does not include the cost of operative procedures.
***Medstat private sector reimbursement figures.

Preventing Limb Loss in the Department of Veterans Affairs

The widespread prevalence of peripheral neuropathy, the 7.5 percent incidence and 28 percent prevalence of prior foot sores, and the frequency of prior revascularization and amputation suggest a large VA population at high risk for diabetic amputation. Available clinical trial data describe several effective foot care interventions. These include a randomized trial showing the effectiveness of outpatient diabetes education in decreasing diabetic lower limb amputations (11), the reduction in foot ulcer healing time with total contact casts (12), and studies that evaluated patient, health care provider, and system interventions related to foot care. The group of patients who were randomized to intervention received patient education and contracts, while their providers received education, guidelines, and prompts. Over the 1-year study, results showed a decreased prevalence of serious foot lesions, increased frequency of foot examinations, and increased podiatry referrals (13).

CONCLUSIONS

The causal pathway data suggest that footwear is an important component cause of amputation and deserves further attention. This finding is consistent with reports from other investigators that indicate that 39–76 percent of amputations in their diabetic clinical and research populations were similarly initiated by ill-fitting footwear (14–16). No experimental trials have been conducted to establish the efficacy of therapeutic footwear in preventing ulcers and/or amputations. Related research includes a cross-over study of patients allocated to intervals of running shoes and their own footwear. Results suggest a reduction in callus formation while wearing the running shoes (17). No conclusions on therapeutic footwear efficacy could be drawn from the Health Care Finance Administration's Therapeutic Shoe Demonstration which reported in 1993, "no evidence that providing therapeutic shoes for diabetics with severe foot disease was not a cost neutral benefit" (18). This study was not designed to test clinical effectiveness of therapeutic footwear and in a subsequent population the authors concluded a more orderly approach to the trial would have been to first determine the conditions for clinical effectiveness before assessing their effects on Medicare costs (19). A descriptive study following ulcerated patients over 2 years found that reulceration occurred in 72 percent of patients who resumed wearing their own footwear compared to 26 percent of patients who continued wearing prescribed footwear (15). In a Swedish cohort study, individuals with prior healed foot ulcers reported 1-, 3-, and 5-year reulceration rates at 34 percent, 61 percent, and 70 percent, respectively (16).

Available therapeutic footwear is limited and patients complain because it is expensive, unattractive, not a covered benefit, and wears out quickly. Shoe requirements based on the foot geometry of diabetic patients with foot insensitivity suggest that many patients belong in therapeutic shoes with extra width across the metatarsal heads and extra depth in the toe box, while only a small percentage of patients have such extensive foot deformities that a custom shoe is required.

In VA, many medically needy diabetic veterans are denied footwear because they do not meet the service connected or other "mandatory care" requirements. Health care organizations such as VA should test footwear in prospective intervention trials to determine the appropriateness of providing this benefit to the high risk service recipients and the achievable cost savings from this action.

REFERENCES

1. Reiber GE, Smith D, McNamara K, Preston S. The epidemiology of amputation in the United States: 1989–1992. In press.
2. US Department of Health and Human Services. Centers for Disease Control, Division of Diabetes Translation. Diabetes surveillance: annual report. Atlanta, GA: DHHS, 1993.
3. Rothman KJ. Causes. Am J Epidemiol 1976:104:587–92.
4. Rothman KJ. Modern epidemiology, Boston, MA: Little Brown, 1986.
5. Pecoraro RE, Reiber GE, Burgess EM. Pathways to diabetic limb amputation: basis for prevention. Diabetes Care 1990:13:513–21.
6. Reiber GE, Pecoraro RE, Koepsell TD. Risk factors for amputation in patients with diabetes mellitus: a case-control study. Ann Int Med 1992:117:97–105.
7. Linstone HA, Turoff M. The Delphi method: techniques and applications. Reading, MA: Addison-Wesley, 1975.
8. VA Cooperative Study in Health Services #7, A comprehensive system for quality improvement in ambulatory care. Hines Center for Cooperative Studies in Health Services, Interim Report 9/94.
9. Medstat Systems DRG guide, Ann Arbor, MI: Medstat, 1993.
10. Wright WE, Kaplan, GA. Trends in lower extremity amputations in California, 1983–1987. California Chronic and Sentinel Diseases Surveillance Program Resource Document, Sacramento, CA: CA Dept. of Health Serv, 1989.
11. Malone JM et al. Prevention of amputation by diabetic education. Am J Surg 1989:158:520–4.

12. Mueller MJ et al. Total contact casting in treatment of diabetic foot ulcers-controlled clinical trial. Diabetes Care 1988:11:111–5.
13. Litzelman DK, et al. Reduction of lower extremity clinical abnormalities in patients with non-insulin dependent diabetes mellitus. Ann Int Med 1993.
14. Assal JP. Educating the diabetic patient: difficulties encountered by patients and health care providers. In: Concepts for the ideal diabetes clinic. Berlin: Gruyter, 1992:73–87.
15. Edmonds ME et al: Improved survival of the diabetic foot: the role of a specialized foot clinic. Q J Med 1986:60:763–71.
16. Apelqvist J, Larsson J, Agardh CD. Long term prognosis for diabetic patients with foot ulcers. J Intern Med 1993:223:485–91.
17. Soulier S. The use of running shoes in the prevention of plantar diabetic ulcers. Am J Pod 1986:76:395–9.
18. Wooldridge J, Bergeron J, Moreno L, Thornton C. Evaluation of the Medicare therapeutic shoe demonstration: final summary report: HCFA, DHHS, Mathematica Policy Research, Inc., Princeton, NJ: DHHS, 1993.
19. Wooldridge J, Moreno L. Evaluation of the costs to Medicare of covering therapeutic shoes for diabetic patients. Diabetes Care 1994:17:541–7.

ABSTRACTS OF RECENT LITERATURE

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Abstracts are drawn primarily from the orthotics and prosthetics literature. Selections of articles were made from these journals:

Archives of Physical Medicine and Rehabilitation
Ergonomics
Journal of Prosthetics and Orthotics
Medizinisch-Orthopädische Technik
Physical Therapy
Prosthetics and Orthotics International

Grip Strength in Different Positions of Elbow and Shoulder. Su C-Y, Lin J-H, Chien T-H, et al. Reprinted from *Arch Phys Med Rehabil* 1994:75:812–815 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

This study investigated the effect of shoulder position on grip strength in 80 men and 80 women. A Jamar dynamometer was used to measure the grip in the four testing positions. The four hand strength tests consisted of three positions in which the elbow was maintained in full extension combined with varying degrees of shoulder flexion (ie, 0°, 90°, and 180°) and of one position in which the elbow was flexed at 90° with the shoulder in 0° of flexion. Only the dominant hand was tested. The highest mean grip measurement was recorded when the shoulder was positioned at 180° of flexion with elbow in full extension; whereas the position of 90° elbow flexion with shoulder in 0° of flexion had the lowest grip strength score. In addition, the grip strength measured with the elbow in extension, regardless of shoulder position (ie, 0°, 90°, and 180° of flexion), was significantly higher than when the elbow was flexed at 90° with the shoulder positioned at 0° of flexion. Finally, grip strength differed significantly for both sexes and for each age group. The grip values of the standardized 90° elbow flexed position were further analyzed to determine the average performances in the study population. For men, grip strength peaked within the 20 to 39 years age group and gradually declined thereafter. For women, the highest mean grip measurement was recorded in the 40- to 49-year-old age group and then deteriorated with age. The findings are valuable in the evaluation and rehabilitation of hand injured patients.

Influence of Prosthetic Foot Design on Sound Limb Loading in Adults with Unilateral Below-Knee Amputations. Powers CM, Torburn L, Perry J, Ayyappa E. Reprinted from *Arch Phys Med Rehabil* 1994:75:825–829 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Altered gait patterns resulting from amputation has been implicated as a possible mechanism for early degenerative changes in the sound limb of unilateral amputees. The purpose of this study was to examine the ground reaction force characteristics and joint motion in this population. Ground reaction forces, joint motion, and stride characteristics of 10 traumatic below-knee amputees were analyzed while wearing five different prosthetic feet (SACH, Flex-foot, Carbon Copy II, Seattle, and Quantum). Subjects used each foot for 1 month prior to testing. Results indicated that the Flex-foot significantly reduced the initial peak of the vertical ground reaction force on the sound limb compared to all other feet tested ($p < 0.0001$). The SACH foot consistently produced the greatest ground reaction forces on the sound limb; however, this was not statistically significant. The effective factor of the Flex-foot appears to be minimization of the center of gravity elevation, which was accomplished through a significant increase in terminal stance dorsiflexion compared to the other feet tested ($p < 0.0001$).

Keramik in der Medizintechnik (Ceramics for Medical Application). Willmann G. Reprinted from *Med Orth Tech* 1994:114:147–148.

Ceramics for medical application (i.e. bioceramics) had been developed 20 years ago. Today the most important application is the ceramic ball head for total hip replacement. Up to now about 2 million femoral ball heads have been used worldwide.

Most Comfortable Listening Level as a Function of Age. Coren S. Reprinted from Ergonomics 1994:37:1269–1274.

The Most Comfortable Listening Level (MCLL) was determined for a running speech signal in a sample of 799 subjects, ranging in age from 17 to 92 years of age. The MCLL increased monotonically with increasing chronological age, in a non-linear relationship. Before the age of 40 MCLL increased approximately one-third of a dB per annum, while after the age of 65 MCLL increased by more than one half dB per year. Over the 75 year age range MCLL rose by 34 dB. Confirming earlier reports, MCLL was also found to be related to hearing sensitivity. Some implications for the design of sound systems are suggested.

A New Modular Six-bar Linkage Trans-femoral Prosthesis for Walking and Squatting. Chakraborty JK, Patil KM. Reprinted from *Prosthet Orthot Int* 1994:18:98-108.

Four-bar linkage mechanisms produced by many designers of knee joints for trans-femoral prostheses can provide knee rotation to permit walking only. In Afro-Asian countries people are accustomed to a squatting posture in their daily activities. A six-bar linkage knee-ankle mechanism trans-femoral prosthesis is described which was developed and fitted to an amputee. The motion patterns of the ankle, knee and thigh during walking and squatting (obtained using a flickering light emitting diode system) for the above prosthesis is compared with motion patterns obtained for normal subjects. The closeness between both the patterns establishes the suitability of the new modular trans-femoral prosthesis for producing near normal patterns of motion during walking and squatting. The additional facility of cross-legged sitting provided in the prosthesis makes it functionally suitable for Afro-Asian amputees.

Physiologic Responses to Forward and Retrograde Simulated Stair Stepping. Ryan PT, Plowman SA, Ball TE, Looney MA. Reprinted from *Arch Phys Med Rehabil* 1994:75:798-802 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

This study compared the physiologic responses to forward and retrograde simulated stair stepping on the StairMaster 4000 PT. Twenty male subjects (mean age 23.65 ± 1.63 years) volunteered for this study. Subjects completed a practice trial of 6 minutes of both forward and retrograde stepping at Level 5. Each experimental trial was divided into four 3-minute stages: Level 3, Level 5, Level 7, and Level 9. Heart rate, blood pressure, and rating of perceived exertion (RPE) were recorded during the second minute of each stage. Expired gases were analyzed and averaged over the last 2 minutes of each stage. Caloric expenditure and delta efficiency were later calculated. Data were analyzed using a 2×4 ANOVA (direction by level) and 2×3 ANOVA (for delta efficiency). Compared to forward responses, retrograde heart rates were significantly higher at Levels 7 and 9 ($p < 0.01$). Retrograde responses for RPE, metabolic equivalents (METS), and caloric expenditure were significantly higher at ($p < 0.01$) Levels 5, 7, and 9 when compared to forward responses. However, the results of this study show that these differences between forward and retrograde stepping are not practically meaningful.

Prediction of Ambulatory Performance Based on Motor Scores Derived from Standards of the American Spinal Injury Association. Waters RL, Adkins R, Yakura J, Vigil D. Reprinted from *Arch Phys Med Rehabil* 1994:75:756-760 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Assessment of strength using motor scores derived from the standards of the American Spinal Injury Association (ASIA) was compared with assessment using motor scores based on biomechanical aspects of walking in the prediction of ambulatory performance. Measurements of strength, gait performance, and the energy expenditure were performed in 36 spinal cord injured patients. The ASIA scoring system compared favorably with the biomechanical scoring system. The ASIA score strongly correlated with the percent increase in the rate of O_2 consumption above normal ($p < .0005$), O_2 cost per meter ($p < .0006$), peak axial load exerted by the arms on crutches ($p < .0001$), velocity ($p < .0001$), and cadence ($p < .0001$). Patients with lower extremity ASIA scores ≤ 20 were limited ambulators with slower average velocities at higher heart rates, greater energy expenditure, and greater peak axial load exerted on assistive devices than patients with lower extremity ASIA scores ≤ 30 who were community ambulators. We conclude the ASIA motor score is a simple clinical measure that strongly correlates with walking ability.

Pressure Ulcers: A Review. Yarkony GM. Reprinted from *Arch Phys Med Rehabil* 1994:75:908-917 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

This article reviews the etiology, pathology, description, risk factors, prevention, medical and surgical management, and complications of pressure ulcers. Pressure ulcers, which develop primarily from pressure and shear, are also known as decubitus ulcers, bed sores, and pressure sores. They continue to occur in hospitals, nursing homes, and among disabled persons in the community. Estimates of the prevalence of pressure ulcers in hospitalized patients range from 3% to 14% and up to 25% in nursing homes. Persons with spinal cord injury and the elderly are two groups at high risk. The most common sites of development are the sacrum, ischium, trochanters, and about the ankles and heels. Areas of ongoing research such as electrical stimulation and growth factors are discussed.

Prosthetic and Orthotic Lab Applications in Medical Imaging Head Immobilization. Pilipuf MN, Berry JM, Goble JC, Kassell NF. Reprinted from *J Prosthet Orthot Int* 1994:6:79-82.

A thermoplastic head immobilization system was designed and fabricated at the University of Virginia Medical Center. The department of neurosurgery and the department of prosthetics and orthotics produced a cost-effective head restraint device that is adaptable to a variety of imaging systems. This article highlights the issues involved in material selection, fabrication and the resulting application in MRI imaging studies.

Subjective Benefits of Energy Storing Prostheses. Alaranta H, Lempinen V-M, Haavisto E, et al. Reprinted from *Prosthet Orthot Int* 1994:18:92-97.

The energy storing (ES) prosthesis has been used in the Prosthetic Foundation's workshop since 1987. Subjective responses from 168 amputees (141 trans-tibial and 27 trans-femoral) fitted with the ES prosthesis were analysed. Ratings were generally favourable in comparison with those for conventional prostheses. The most pronounced advantages of the new prosthesis as shown by the ratings were in walking uphill or swift walking. The younger amputees had more benefit than the older ones. High body weight decreased the benefit of the ES prosthesis. The ES prosthesis does not seem to provide any major advantage for the less active amputee whose movements are mainly indoors.

Temporal, Kinematic, and Kinetic Variables Related to Gait Speed in Subjects with Hemiplegia: A Regression Approach. Olney SJ, Griffin MP, McBride ID. Reprinted from *Phys Ther* 1994:74:872-885.

Background and purpose. The gait speed that a patient selects is a well known indicator of overall gait performance. The purpose of this study was to use multiple linear regression to assess the strength of association of temporal, kinematic, and kinetic gait variables with high walking speeds in patients with hemiplegia. **Subjects.** Thirty-two subjects (20 male, 12 female) with an average age of 61 years took part in a saggital-plane gait study of both sides of the body. **Methods.** Data from cinematographic film and a force plate obtained during multiple walking trials were used in a seven-segment link-segment kinetic model of the walking subject to yield temporal, kinematic, and kinetic variables. **Results.** Variables correlating significantly with self-selected speed included the maximum hip extension angle and the maximum hip flexion moment on the affected side, and the maximum ankle and hip powers on both sides. A stepwise regression identified variables most useful in predicting stride speed. For the affected side these variables were the hip flexion moment, the ankle moment range, the knee moment range, and the proportion of double support. Together these variables explained 94% of the variation in gait speed. On the unaffected side, the variables were the percentage of stance phase, the maximum ankle power (push-off), and the maximum hip power (pull-off). They explained 92% of the variation in gait speed. **Conclusion and Discussion.** These results suggest that experimental studies are needed to assess the effects of treatment aimed at increasing ankle power and hip power and at decreasing the stance time on the affected side, and that these studies should be directed at obtaining a larger hip flexion moment and a larger ankle moment range on the unaffected side.

Validation of the Quick Cognitive Screening Test. Mate-Kole CC, Major A, Lenzer I, Connolly JF. Reprinted from *Arch Phys Med Rehabil* 1994:75:867-875 (©1994 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation).

Reports in the literature suggest that 70% to 80% of cognitive deficits with probable organic basis are undetected in patients at risk, especially if symptoms are minimal. Studies have shown that clinically convenient bedside screening tests show low sensitivity with a high rate of false-negative results. The purpose of this study was to validate a sensitive mid-range cognitive screening test to detect cognitive deficits, including those not usually identified by bedside mental status examinations. The Quick Cognitive Screening Test (QCST) was initially developed and subsequently adapted from original unpublished work by the late John McFie. The test was designed to detect global cognitive dysfunction and specific areas of cognitive dysfunction. Areas assessed included orientation, attention and concentration, memory, language, construction, perception, spatial ability, and abstract reasoning. Scoring is multidimensional, each subtest having a score, plus summary and global scores. Thirty-eight neurological patients with a cerebrovascular accident, traumatic brain injury, and other miscellaneous diagnoses, were recruited from a tertiary care center for physical rehabilitation. Fifteen residents from a psychiatric rehabilitation center were also recruited. Thirty-two healthy volunteers from the community served as age-matched controls. The Wechsler Adult Intelligence Scale-Revised (WAIS-R) was used to establish reliability and validity of the QCST. The National Adult Reading Test (NART) was used to estimate premorbid intellectual functioning. Results showed that the QCST identified cognitive impairment in all of the neurological and psychiatric patients assessed. One way analyses of variance of five summary scores showed significant differences between the groups. Scheffe's post hoc analyses ($p < .05$) revealed that the QCST differentiated between the control group and both the psychiatric and neurological groups. There were significant correlations (Pearson r) between the QCST Global Score and the WAIS-R FSIQ, the QCST Verbal Score and the WAIS-R VIQ, and the QCST Nonverbal Score and WAIS-R PIQ, confirming the validity of the QCST. Additionally, there were significant correlations between the QCST Global Score and the NART estimated FSIQ, the QCST Verbal Score and the NART estimated VIQ, and the QCST Nonverbal Score and the estimated NART PIQ. Reliability was determined by the alpha coefficient (.87). The QCST shows promise as a brief, reliable, and valid screening instrument for detection of cognitive dysfunction in neurological patients.

Validity and Reliability of a New Assessment of Lower-extremity Dysfunction. Oberg U, Oberg B, Oberg T. Reprinted from *Phys Ther* 1994:74:861-871.

Background and Purpose. A 20-variable assessment system for evaluation of lower-extremity dysfunction has been constructed with special consideration of the needs of the physical therapist. The variables are classified into five subgroups: hip impairment, knee impairment, physical disability, social disability, and pain. **Subjects.** One hundred five patients with osteoarthrosis of the hip and knee, all accepted for total joint replacement arthroplasty, were tested. The mean age of the patients was 69 years (SD = 9.0, range = 46–91). **Methods.** The original grouping of the variables was analyzed for content validity with a factor analysis. The results from a subgroup of 42 patients were tested for intertester reliability with the Goodman-Kruskal gamma coefficient. **Results.** The factor analysis indicated a factor solution consistent with the primary grouping except for two variables. The correlation between two independent physical therapists was .99 to 1.00 for different variables, indicating excellent intertester reliability. **Conclusion and Discussion.** In the author's opinion, the new assessment system provides a reasonably valid, reliable, inexpensive, and easy-to-use measurement and fulfills the needs Of the physical therapist for functional evaluation of the lower extremity.

Die Achillodyn-Orthese im Therapieplan van Achillessehnenerkrankungen (The Achillodyn Orthosis in the Therapy of Achilles Tendon Disorders). Wiseman J, Hildebrandt HD. Reprinted from *Med Orth Tech* 1994:114:142-144.

Pain in the area of the Achilles tendon is usually caused by overloading and wrongly loading. This structure is seen often in sport. The Achillodyne orthosis is used in the conservative treatment of acute and chronic paratenonitis achilleae, achillodynia as well as painful irritation following surgery of the Achilles tendon. Together with antiplogistic, physiotherapeutic and electrophysical measures this device has been successfully employed in the therapy scheme. Through slight raising of the heels (8 mm heel wedge on both sides), the Achilles tendon becomes stress-relieved. The adjustable paratendinous silicon supports have a mild massaging effect through function. The ankle supports give the patient a feeling of more security. If necessary, the plantar heel wedge can be easily removed.

The advantage of this orthosis lies in its special action when used properly.

Amputation and Reflex Sympathetic Dystrophy. Geertzen JHB, Eisma WH. Reprinted from *Prosthet Orthot Int* 1994:18:109-111.

Reflex sympathetic dystrophy is a chronic pain syndrome characterized by chronic burning pain, restricted range of motion, oedema and vasolability. Patients are difficult to treat and the prognosis is very often poor. This report emphasizes that an amputation in case of a reflex sympathetic dystrophy is mostly due to a too late recognition of this syndrome. In the international literature little is written about an amputation as a therapy for reflex sympathetic dystrophy. It is only mentioned as a therapy in the end stages of this syndrome. Sometimes a rejected amputation, as in this case report, can have a relatively good result. An early recognition of this pain syndrome produces the best possible outcome.

A Body Powered Prehensor with Variable Mechanical Advantage. Frey DD, Carlson LE. Reprinted from *Prosthet Orthot Int* 1994:18:118-123.

The purpose of this research was to improve body powered, voluntary closing (VC) prosthetic prehension. A prototype prehensor with variable mechanical advantage was fabricated and tested. The device operates at low mechanical advantage during sizing of an object to reduce cable excursion requirements. It shifts to high mechanical advantage during gripping to allow high prehensile forces to be generated with reduced cable tension. The prototype provides a mechanical advantage of 2.4, nearly five times that of conventional VC devices. The prototype also acts as a holding assist, after grip forces are applied, they can be maintained with a cable tension of only 3 lb (13.34N). Field testing indicated that the device performs well in many tasks. The mechanism allows greater range of motion while an object is grasped than standard voluntary closing prehensors. However, the device performed poorly in grasping very compliant objects. To address this problem, a switch has been incorporated into the prototype to allow it to be used in a free-wheel mode.

A CAD Analysis Programme for Prosthetics and Orthotics. LeMaire E. Reprinted from *Prosthet Orthot Int* 1994:18:112-117.

A CAD (computer aided design) analysis software package (CADVIEW) was designed for use with prosthetic and orthotic CAD CAM (computer aided design/computer aided manufacture) systems. Using the Microsoft Windows 3.1 environment, CADVIEW provides a series of anatomical shape viewing and analysis tools. These tools include simultaneous display of multiple sockets and multiple views, two dimensional (2D) and three dimensional (3D) measurement, shape statistics, multi-shape alignment, cross-sectional comparison, colour coded 3D comparison, resolution enhancement, and image copying capabilities. This programme should be of benefit to clinicians and researchers who wish to assess and/or compare CAD data generated by MS-DOS based CAD CAM systems.

A Comparison of Gait Characteristics in Young and Old Subjects. Ostrosky KM, VanSwearingen JM, Burdett RG, Gee Z. Reprinted from *Phys Ther* 1994:74:637-646.

Background and Purpose. The purpose of this study was to describe and compare active range of motion during free-speed gait in younger and older people. **Subjects.** Sixty volunteers in good health were studied. Thirty subjects (15 male, 15 female) were between 20 and 40 years of age, and 30 subjects (15 male, 15 female) were between 60 and 80 years of age. **Method.** Subjects were videotaped walking down a 6-m walkway) with reflective markers at six locations along their right side. The videotape was analyzed for nine gait characteristics using a two-dimensional video motion analysis system. Differences in gait characteristics between the two groups were examined using a multivariate analysis of variance, followed by univariate F tests. **Results.** Two gait variables—knee extension and stride length—were significantly different between groups, and differences in velocity approached significance. **Conclusion and Discussion.** For individuals in good health, the gait of older people differs from the walking pattern of young people for selected variables. Older people demonstrate less knee extension and a shorter stride length compared with younger people. Differences in self-paced walking velocity between old and young people may have influenced the gait characteristics measured.

Comparison of Ultrasound/Ultraviolet-C and Laser for Treatment of Pressure Ulcers in Patients with Spinal Cord Injury. Nussbaum EI, Blemann I, Mustard B. Reprinted from *Phys Ther* 1994:74:812-825.

Background and Purpose. The purpose of this study was to compare in patients with spinal cord injury the effect on wound healing of nursing care alone with the effect on wound healing of nursing care combined with either laser treatment or a regimen of ultrasound and ultraviolet-C (US/UVC). **Subjects** Twenty patients (22 wounds) were randomly assigned to the treatment groups. **Methods.** All patients received standard wound care consisting of wound cleaning twice daily, application of moist dressings, and continuous relief of pressure until the wounds were healed. The laser protocol consisted of three treatments weekly using a cluster probe with an 820-nm laser diode and 30 superluminous diodes (10 each at 660, 880, and 950 nm), an energy density of 4j/cm^2, and a pulse repetition rate of 5,000 pulses per second. The US/UVC regimen consisted of five treatments weekly, alternating the treatment modality daily. The pulsed US was applied at a frequency of 3 MHz and a spaced average-temporal average intensity of 0.2 W/cm^2 (1:4 pulse ratio) for 5 minutes per 5 cm^2 of wound area. The UVC dosage (95% emission at 250 nm) was calculated each session according to wound appearance. The dosage level was E_1 for clean/granulating areas, E_3 for purulent/slow-granulating areas; E_4 for heavily infected areas and $2E_4$ for wound debridement. Wounds were traced every 14 days; and surface areas were calculated using the Sigma-Scan Measurement System. Weekly percentage changes in wound area were compared. **Results.** Results showed that US/UVC treatment had a greater effect on wound healing than did nursing care, either alone or combined with laser. **Conclusion and Discussion.** Ultrasound/ultraviolet-C may decrease healing time and may allow faster return to rehabilitation programs, work, and leisure activities for patients with spinal cord injury who have pressure ulcers.

Early Management of Elderly Dysvascular Below-Knee Amputees. Cutson TM, Bongiorni D, Michael JW, Kochersberger. Reprinted from *J Prosthet Orthot* 1994:6:62-66.

The majority of transtibial (below-knee) amputations occur in elderly patients with systemic vascular disease. Rehabilitation efforts toward prosthetic ambulation are frequently delayed awaiting postoperative healing of the vascular compromised limb. Rehabilitation becomes more difficult, more costly and less successful the longer it is delayed after surgery, especially among elderly amputees. Early ambulation reduces the risk of complications such as thromboembolism, pneumonia and deconditioning in the older patient as well as enhances remaining life. An early coordinated post-amputation rehabilitation program reduces the time to prosthetic ambulation and the risk of further debility and failure among elderly amputees. The rigid removable dressing was incorporated into the program and found to be a safe method of residual limb shrinkage among elderly dysvascular transtibial amputees.

The Effect of Semirigid Dressings on Below-Knee Amputations. MacLean N, Fick GH. Reprinted from *Phys Ther* 1994:74:668-673.

Background and Purpose. The effect of using semirigid dressings (SRDs) on the residual limb of individuals who have had below-knee amputation as a consequence of peripheral vascular disease was investigated, with the primary question being: Does the time to readiness for prosthetic fitting for patients treated with the SRDs differ from that of patients treated with soft dressings? **Subjects.** Forty patients entered the study and were alternately assigned to one of two groups. Nineteen patients were assigned to the SRD group, and 21 patients were assigned to the soft dressing group. **Methods.** The time from surgery to readiness for prosthetic fitting was recorded for each patient. Kaplan-Meier survival curves were generated for each group, and the results were analyzed with the log-rank test. **Results.** There was a difference between the two curves, and an examination of the curves suggests that the expected time to readiness for prosthetic fitting for patients treated with the SRDs would be less than half that of patients treated with soft dressings. **Conclusion and Discussion.** The results suggest that a patient may be ready for prosthetic fitting sooner if treated with SRDs instead of soft dressings.

The Effect of Soft Foot Orthotics on Three-dimensional Lower-limb Kinematics during Walking and Running. Eng JJ, Pierrynowski MR. Reprinted from *Phys Ther* 1994:74:836-844.

Background and Purpose. Although foot orthotics are often prescribed to alter the lower-extremity mechanics during the stance period of gait, there is little documentation of the actual effect of foot orthotics on the movement of the lower-extremity joints during walking and running. This study examined the effect of foot orthotics on the range of motion of the talocrural/subtalar joint and the knee joint in three dimensions during walking and running. **Subjects.** Ten female adolescent subjects, aged 13 to 17

years ($\bar{X}$ = 14.4, SD = 1.1) who were diagnosed with patellofemoral pain syndrome and exhibited forefoot varus greater than 6 degrees and/or calcaneal valgus greater than 6 degrees participated in the study. **Methods.** Thirty strides of walking and running on a treadmill were recorded for each of the orthotic and nonorthotic conditions for each subject using an optoelectronic recording technique. Analyses of variance for repeated measures were performed on the range of motion of the talocrural/subtalar joint and knee joint for each plane of motion (ie, six separate analyses). The main factors of each analysis were the effect of the orthotic (orthotic condition versus nonorthotic condition), mode of ambulation (walking and running), and phase of the stance period (contact, mid-stance, and propulsion). **Results.** No differences were found in sagittal-plane movements. Reductions of 1 to 3 degrees occurred with orthotic use for the talocrural/subtalar joint during walking and running in the frontal and transverse planes. The orthotics reduced knee motion in the frontal plane during the contact and mid-stance phases of walking, but increased the motion during the contact and mid-stance phases of running. **Conclusions and Discussion.** The study shows that corrections to the static position of forefoot varus and calcaneal valgus can result in changes in transverse- and frontal-plane motion of the foot and knee during walking and running.

Effects of Two Types of Chairs on Stature Change and Comfort for Individuals with Healthy and Herniated Discs. Michel DP, Helander MG. Reprinted from *Ergonomics* 1994:37:1231-1244.

The objective of this study was to determine if stature change and perceived comfort are significantly different for individuals with either healthy or herniated discs when seated in a conventional chair or a sit-stand chair. Sixteen subjects were studied (5 young/healthy, 6 old/healthy, 5 old/herniated). Subjects performed a search task on a computer screen during two 2 h sessions for two consecutive days, with a different chair each day. Changes in stature were measured with a stadiometer. General comfort and body parts discomfort rating scales were administered every 30 min. The main findings were: (1) for all subjects, the sit-stand chair produces less height loss than the conventional chair; and (2) for both chairs, subjects with herniated discs lost more height than subjects with healthy discs. A positive correlation was observed for height loss and age with the sit-stand chair. Subjects with herniated discs felt relatively more uncomfortable in the conventional chair and more comfortable in the sit-stand chair compared to subjects with healthy discs. Both old healthy and young healthy subjects felt more comfortable in the conventional chair in comparison to the sit-stand chair. But, old healthy subjects had a relatively greater perception of comfort in the conventional chair compared with the younger subjects.

Energy Expenditure of Trans-tibial Amputees during Ambulation at Self-selected Pace. Gailey RS, Wenger MA, Raya M, et al. Reprinted from *Prosthet Orthot Int* 1994:18:84-91.

The purpose of this investigation was two-fold: 1) to compare the metabolic cost (VO_2), heart rate (HR), and self-selected speed of ambulation of trans-tibial amputees (TTAs) with those of non-amputee subjects; and 2) to determine whether a correlation exists between either stump length or prosthesis mass and the energy cost of ambulation at the self-selected ambulation pace of TTAs. Subjects were thirty-nine healthy male non-vascular TTAs between the ages of 22 and 75 years (mean $\pm$ sd = 47 $\pm$ 16). All had regularly used their prosthesis for longer than six months and were independent of assistive ambulation devices. Twenty-one healthy non-amputee males aged 27–47 years (31 $\pm$ 6) served as controls. Subjects ambulated at a self-selected pace over an indoor course, with steady-state $V0_2$, HR, and ambulation speed averaged across minutes seven, eight and nine of walking. Results showed that HR and $V0_2$ for TTAs were 16% greater, and the ambulation pace 11% slower than the non-amputee controls. Significant correlations were not observed between stump length or prosthesis mass and the energy cost of ambulation. However, when the TTA subject pool was stratified on the basis of long and short stump length, the former sustained significantly lower steady-state VO_2 and HR than the latter while walking at comparable pace. These data indicate that stump length may influence the metabolic cost of ambulation in TTAS.

Experiences with Respect to the ICEROSS System for Trans-tibial Prostheses. Cluitmans JI, Geboers M, Deckers J, Rings F. Reprinted from *Prosthet Orthot Int* 1994:18:78-83.

This article describes the authors' initial experiences and those of their patients with respect to the ICEROSS system for trans-tibial prostheses. Up to October 1992, 54 patients attending the ''Hoensbroeck'' Rehabilitation Centre received such a prosthesis.

With the aid of patients' records an all-round evaluation has been made. In addition, a survey was undertaken and an examination made amongst the 43 patients who responded to a written request. For 26 patients who were provided with the ICEROSS as a second appliance after having used an older kind of prosthesis a comparison was made with the old system. In general these patients considered the new prosthesis as providing a clear improvement.

Gait in Male Trans-tibial Amputees: A Comparative Study with Healthy Subjects in Relation to Walking Speed. Hermodsson Y, Ekdahl C, Persson BM, Roxendal G. Reprinted from Prosthet Orthot Int 1994:18:68-77.

Walking speed, stance duration and ground reaction forces were studied with the use of a stable force platform (Kistler) in 24 male trans-tibial amputees and 12 healthy subjects matched for sex and age. The aim of the study was to compare the gait performance of two groups with unilateral trans-tibial amputations for either vascular disease or trauma and also to compare the results of the two groups with the results of a group of healthy subjects. Multiple linear regression analysis was used to compare the stance duration and the ground reaction forces in relation to walking speed. The vascular and traumatic amputees had significantly reduced walking speeds compared with the healthy subjects, 0.85 ± 0.2 m/s and 0.99 ± 0.2 m/s, respectively, as compared to 1.42 ± 0.2 m/s. By comparing the vascular and traumatic amputees with the healthy subjects in relation to walking speed, it was shown that the gait performance of the vascular amputee differed from that of the traumatic amputee, a difference that was not caused by the reduced walking speed. The active forces during push off on both the healthy ($p = 0.02$) and the prosthetic leg ($p = 0.003$) in the trauma group were not found in the vascular group. This disparity could be an effect of the systemic disease. It may be argued that the results of this study contribute to the understanding of the reduced walking ability of the vascular amputee and should be borne in mind when planning rehabilitation.

BOOK REVIEWS

by
Joan E. Edelstein, MA, PT; Catherine W. Britell, MD; Roger M. Glaser, PhD; Gregory L. Goodrich, PhD; Mary E. Cupo, BS, KT

Orthopaedic Physical Therapy, second edition. Edited by Robert Donatelli and Michael J. Wooden. New York: Churchill Livingstone, 1994. 778 pp, cloth, illustrated, price not available.

by Joan E. Edelstein, MA, PT
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Many individuals treated by rehabilitation specialists have orthopedic disorders either as the primary problem or as a concurrent or subsequent accompaniment to other impairments. Robert Donatelli, PhD, PT, OCS and Michael J. Wooden, MS, PT, OCS have edited a substantial textbook incorporating the contributions of 31 colleagues, most of whom are physical therapists who are orthopedic clinical specialists; orthopedists, podiatrists, and an occupational therapist also have written chapters.

A brief first section concerns fundamental principles of orthopedic treatment. Characteristics and responses of normal and traumatized muscle, connective tissue, and nerve are reviewed. Of particular interest are the devastating effects of immobilization on these structures. Bed rest also has deleterious effects on the cardiopulmonary and peripheral vascular systems.

The heart of the book lies in the second and third sections devoted to the upper and lower quarters of the body, respectively. Evaluation of the upper quarter is based on the functional anatomy of the shoulder girdle and cervical and thoracic spine. Normal and abnormal shoulder and cervical postures are related to various entrapment neuropathies. The review of patterns of pain referral should enable clinicians to determine the site of pathology. Use of a comprehensive screening examination is illustrated with a case study. The influence of the neck on the temporomandibular joint is explored. In addition to the overall evaluation of the upper quarter, the section has a chapter on the evaluation and treatment of the cervical spine and thoracic outlet with a good review of anatomy and numerous diagnostic tests.

Treatment may involve heat, cold, traction, soft tissue and joint mobilization, therapeutic exercise, and supports. Collars are recommended only for acute conditions and segmental instability. Robert Sprague differentiates among the mechanisms by which spinal manipulation relieves pain and increases flexibility of the neck. Various types of shoulder dysfunction are addressed, primarily from the point of view of physical and arthroscopic evaluation. Elbow injuries (especially those incurred in athletics) and wrist and hand dysfunctions are also discussed, primarily their anatomy and assessment, with a chapter on reconstructive surgery of the hand, including amputation of one or more digits. Wooden's chapter on upper extremity mobilization is superbly detailed.

The lower quarter receives comparable attention, with a good functional anatomy and kinesiologic introduction to examination of the patient. The chapter on gait presents kinematics and kinetics in terms of closed chain motion with particular emphasis on the foot. Motion and muscle activity at the ankle, knee, and hip are included, and the ground reaction forces are mentioned. Special attributes of running are delineated. Other chapters in the third section relate to the lumbar spine, hip, knee, foot, and ankle. Trunk orthoses receive scant attention, although acknowledged to relieve pain.

The final section pertains to special considerations, namely a relatively brief discussion of orthopedic problems in the patient with neuropathy and a comprehensive presentation of soft tissue mobilization.

This massive textbook exhibits the advantages and drawbacks evident in most books created by numerous contributors. Most chapters are lavishly illustrated with diagrams and excellent photographs. Many techniques are shown in step-by-step fashion with a series of photographs which may guide clinicians. It is apparent that many procedures can be injurious if done incorrectly; consequently, the text should be considered a supplement to supervised instruction. Tables, evaluation protocols, case studies, and an extensive list of current references at the end of each chapter increase the book's utility. The various authors place differing degrees of emphasis on the amount of anatomy and treatment specificity and rationale included in the book. The role of orthoses in the management of upper and lower limb and trunk disorders is virtually ignored.

Orthopedic Physical Therapy is an excellent textbook for student physical therapists who have the benefit of

professorial guidance. Specialists in rehabilitation will be particularly interested in the material relating to functional anatomy and patient examination.

Clinical Assessment of Muscle Function with a Computer-Assisted Hand-Held Dynamometer. Maria Eugien Roebroeck. Amsterdam: Proefschrift Vrieje Universiteit Amsterdan

by Catherine W. Britell, MD

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This is a doctoral thesis which is comprised of four original papers and discussion by the author on the subject of using a computer-assisted hand-held dynanometer for static and dynamic strength determination for flexion and extension of the knee. The introduction is a nice review of the history and biomechanical principles of various methods of muscle testing. The first and second papers review reliability assessment measures and their applicability to the issue of muscle testing and apply them to this method. The third paper is a study of the blomechanics of sit-to-stand transfers, documenting joint motion and muscle function for this activity. The fourth paper attempts to correlate force measurements on the muscles acting on the knee using a computer-assisted hand-held dynamometer with ability measures. The thesis concludes with a discussion of the practical applicability and generalization of these methods and suggests future research into the clinical uses of such a device. The thesis comprehensively explores and documents many aspects of hand-held dynamometry in functional testing of muscles. It may be a useful as a starting resource for researchers exploring the use of this methodology for clinical investigation, and it is a viable addition to the literature in this area. It has little applicability to clinical practice of rehabilitation or assistive technology service delivery at this time.

Physical Strain and Physical Capacity in Men with Spinal Cord Injuries. Thomas Janssen. Vrije Universiteit Amsterdam: Academisch Proefschrift, 1994, 162 pp.

by Roger M. Glaser, PhD

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This book represents the doctoral dissertation research conducted by Thomas Janssen at Vrije Universiteit in Amsterdam, The Netherlands. It is a comprehensive report from studies conducted on SCI men who use manual wheelchairs and concerns the physical strain they experience during specific activities of daily living, their physical capacity and longitudinal changes over a 3-year period, and their CHD risk in relation to the able-bodied population. Both the methodology used (Chapters 2–5) and data that address the above problems (Chapters 6–9) are provided.

The main questions addressed are:

1. What is the influence of various wheelchair exercise stress test protocols on the determination of physical capacity?
2. Can the physical capacity of persons with SCI be readily determined via graded exercise stress tests?
3. Are selected parameters of physical capacity interrelated in persons with SCI?
4. Can the physical strain of performing standardized ADL tasks be readily determined?
5. What is the physical strain experienced during normal daily life in relation to lesion level in wheelchair users with long-standing SCI?
6. Is the physical strain experienced during standardized ADL tasks directly related to lesion level, and is it inversely related to the physical capacity?
7. Can longitudinal changes in physical capacity be observed, and do these changes coincide inversely with changes in physical strain for performing tasks?
8. Which are the most important determinants (physical capacity, personal factors, behavioral factors) of CHD risk indicators (lipid profile and blood pressure) in the SCI men tested, and do they have inferior levels of CHD risk indicators than able-bodied men?

This book is a valuable resource for those working in the field of SCI rehabilitation and for those performing research on problems experienced by wheelchair users during daily tasks. It provides unique data that are currently scarce, and stimulates thought about how to study these problems in order to determine potential solutions.

Research Abstracts on Blindness in India. Edited by Subash Datrange and Jayasree Mokkapati. Bombay, India: NAB Louis Braille Memorial Research Centre, 1993. Price not available.

by Gregory L. Goodrich, PhD

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Research Abstracts on Blindness in India is an unusual book in that it consists entirely of abstracts of papers

published in India on the general topics of blindness and visual impairment. It is of some interest precisely because of its unique nature.

The abstracts included were published between 1944 and 1993, and they form a capsule view of the development of the "blindness field" in one country. The contents are a chronicle of an interesting body of work as it evolved in one of the most populous countries of the world. Historians of the field will find it of interest, as will other scholars seeking a glimpse of the development of a social service field in India.

All citations in the book include the title, author's name(s), and (for most) date, journal or place of publication. The authors state in their preface that the primary criteria for inclusion was the "availability of the original articles/papers/dissertations." Many articles do indeed include sufficient detail to expect that an interested person could locate the original; however, the editors might have saved the reader a good deal of effort by clearly indicating how to obtain copies of manuscripts. Some abstracts are for papers presented at conferences or published locally by an agency or institute and may be difficult for people outside India to find.

The topics presented in the 117 abstracts included in the book are wide ranging. Samples areas include: educational programs, assessment of blind and visually impaired students, employment and vocational rehabilitation, psychosocial factors, Braille, socioeconomic conditions, demographics, and services. At the end of the book, the editors included title, author, and key word indexes for each of the 117 abstracts included in the book. The indexes are of some importance to the reader since the editors did not arrange the abstracts in any particular order.

Tallying the dates of publication for the abstracts yielded a coherent trend. Most of the abstracts (about 70 percent) were written between 1980 and 1989. About 16 percent were done between 1970 and 1979, and about 11 percent were done between 1960 and 1969. Only two abstracts were written prior to 1960, and about 10 percent were written between 1990 and 1993. This clearly shows that professional interest concerning blindness and visual impairment is growing in India.

The quality of the abstracts varies considerably, no doubt in large part due to translation difficulties. Still, there are important issues explored in the pages of this book. For example, one abstract, "The use of media materials and role training programs with parents in India" by M.L. Mathur, deals with a topic familiar to any special education teacher anywhere in the world. "Parents are generally conscious of the role they are expected to play in bringing up their normal children, but they do not know what role to assume while raising a child with visual impairment." The reader of this book will find that much of its value is the cross cultural perspective it provides on problems common to all societies.

Living in the State of Stuck: How Technology Impacts the Lives of People with Disabilities. Marcia J. Scherer. Cambridge, MA: Brookline Books, 1993. 189 pp, cloth $32.95, paper $24.95

by Mary E. Cupo, BS, KT
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Living in the State of Stuck provides true, personal insights into the differing effects technology can have on the quality of life for persons with disabilities. Senator Tom Harkins states in his Foreword, "Marcia Scherer persuasively reminds us that as technology is used to increase opportunities for individuals with disabilities to become productive members of society, the needs of the whole person must be taken into account," and this statement sets the tone for the text. Provision of assistive technology alone does not automatically mean that we have improved the quality of life for that individual. There is much more. Scherer's thesis is that we must treat the whole person and not just his or her nonfunctioning mechanical parts. As Harkins puts it, "Services that address physical impairment but the emotional, personal, and social needs, must be included or else the true benefits of technology are lost."

Peter Axelson, President of Beneficial Designs, Inc., notes in his Introduction that "this book has something unique to offer to both the assistive technology expert or those who may be new to the field." Unfortunately, the attitudes and personal feelings of persons with disabilities oftentimes get neglected when it comes to the provision of assistive devices. How many times have we experienced the lack of compliance on the part of a user towards a "new and improved" device that we know will make life easier? The important issue here is to allow the user to become an active participant in the assistive technology process. Increased compliance will ensure maximum benefit for the person with disabilities. As Axelson writes, "the technology, however helpful it may be in its design, cannot function without acceptance from potential users."

The life stories related in Scherer's book serve as crucial reminders that "technology alone is rarely the answer to a

person's enhanced quality of life." Each individual must be given the opportunity to exercise and develop not only his or her physical abilities but also, as sociologist Irving Zola points out, "to exercise self-directed choices, manage feelings, request assistance with dignity, refuse it with diplomacy, [and] learn to be assertive in job maintenance and advancement and in the establishment of new relationships."

Chapter Seven outlines the dilemmas, challenges and opportunities involved in the assistive technology process including problematic issues concerning research on technology use. The material in this chapter offers the reader solid information in realizing that the purpose of an assistive technology is to enhance a person's functioning, esteem, and quality of life. Funding of assistive technologies is discussed along with cost effectiveness (i.e., rehabilitation versus maintenance issues). The author presents many important questions that must be answered when recommending a device for a person. A "consumer-driven" approach will best address that person's unique and challenging needs and empower him or her to attain maximum independence. In the Summary and Reference Notes sections of this chapter, various models and assessment/screening instruments for matching persons with technology are described and should prove useful to anyone involved in the assistive technology field.

The concluding chapter on Future Directions provides an excellent outlook on how "rehabilitation professionals, whether counselors, physicians, nurses, engineers, [or] therapists, must increase their focus on the psychosocial aspects of individual's quality of life." In times where technology proliferates throughout our entire society, the importance of the "human touch" must not be forgotten. Scherer goes on to report "the persons who have shared their experiences in this book said that, despite efforts [such] as the UN Decade of Disabled Persons (which ended in 1992), the Americans With Disability Act, and their own accomplishments, they still feel inadequate, like marginal participants in society." Interestingly enough, this sentiment could probably be shared by persons with or without disabilities. The author provides the reader with a personalized view of the world as experienced by several persons with disabilities. The book presents information that would be helpful to both the seasoned professional and someone new to the rehabilitation field. The ideas and material may help bring the focus back to the user and with that, hopefully, more effective application of assistive technology and successful rehabilitation outcomes may result.

PUBLICATIONS OF INTEREST

This list of references offers *Journal* readers significant information on the availability of recent rehabilitation literature in various scientific, engineering, and clinical fields. The *Journal* provides this service in an effort to fill the need for a comprehensive and interdisciplinary indexing source for rehabilitation literature.

All entries are numbered so that multidisciplinary publications may be cross-referenced. They are indicated as *See also* at the end of the categories where applicable. A listing of the periodicals reviewed follows the references. In addition to the periodicals covered regularly, other publications will be included when determined to be of special interest to the rehabilitation community. To obtain reprints of a particular article or report, direct your request to the appropriate contact source listed in each citation.

AMPUTATIONS

1. Effect of Semirigid Dressings on Below-Knee Amputations. MacLean N, Fick GH, *Phys Ther* 74(7):668-673, 1994.
Contact: Nancy MacLean, DipPT, Dept. of Physical Medicine and Rehabilitation, Foothills Hospital, 1403-29 St. NW, Calgary, AB T2N 2T9, Canada

2. Major Replantation: A Report of Three Cases. Fahmy HWM, Aldanaf AA, *Egyptian Orthop J* 28(3):123-132, 1994.
Contact: Dr. Hani W. Mansour Fahmy, Hand and Upper Limb Centre and Institute, Dokki, Giza

3. Upper Limb Replantation—How I do It. Holmberg J, *Acta Orthop Scand* 65(3):359-360, 1994.
Contact: Jan Holmberg, Dept. of Hand Surgery, General Hospital, S-214 01 Malmo, Sweden

BIOENGINEERING

4. Multireference Apaptive Noise Cancellation Applied to Somatosensory Evoked Potentials. Parsa V, Parker PA, *IEEE Trans Biomed Eng* 41(8):792-800, 1994.
Contact: Vijay Parsa, Institute of Biomedical Engineering, University of New Brunswick, Fredericton, NB E3B 5A3 Canada

BIOMATERIALS

10. Ceramics and Joint Reconstruction. Skinner HB, *Tech Orthop* 8(4):275-279, 1994.
Contact: Harry B. Skinner, MD, PhD, UCSF Medical Center, 500 Parnassus Ave. (MU-320W), San Francisco, CA 94143-0728

BIOMECHANICS

5. Biomechanics and Muscular Activity During Sit-to-Stand Transfer. Roebroeck ME, et al., *Clin Biomech* 9(4):235-244, 1994.
Contact: Dr. M. Roebroeck, Dept. of Rehabilitation Medicine, Free University Hospital, PO Box 7057, 1007 MB Amsterdam, The Netherlands

6. Biomechanics of Malalignment. Chao EYS, et al., *Orthop Clin North Am* 25(3):379-386, 1994.
Contact: Edmund Y.S. Chao, PhD, Orthopaedic Biomechanics Laboratory, School of Medicine, The Johns Hopkins University of Medicine, Ross Research Bldg., Room 225, 720 Rutland Ave., Baltimore, MD 21205-2196

7. Dynamics of Knee Malalignment. Andriacchi TP, *Orthop Clin North Am* 25(3):395-403, 1994.
Contact: Thomas P. Andriacchi, PhD, Dept. of Orthopedic Surgery, Rush-Presbyterian-St. Luke's Medical Center, 1653 West Congress Pkwy., Chicago, IL 60612

8. Foot Inversion-Eversion and Knee Kinematics During Walking. Lafortune MA, et al., *J Orthop Res* 12(3):412-420, 1994.
Contact: Dr. M.A. Lafortune, School of Human Biology, University of Guelph, Guelph, ON N1G 2W1 Canada

9. Relationships Between Alignment, Kinematic and Kinetic Measures of the Knee of Normal Elderly Subjects in Level Walking. Wang H, Olney SJ, *Clin Biomech* 9(4):245-252, 1994.
Contact: Sandra J. Olney, School of Rehabilitation Therapy, Queen's University, Kingston, ON K7L 3N6 Canada

COMMUNICATION AIDS—HEARING

11. Aging, Hearing Loss, and Hearing Aids: Myths Revisited. Thompson ME, *Am Rehabil* 19(4):20-24, 1994.
Contact: Maureen E. Thompson, MA, Audiology Services Branch, American Speech-Language-Hearing Assoc., Rockville, MD

12. Effects of Age and Hearing Loss on Tasks of Perception and Production of Intonation. Most T, Frank Y, *Volta Rev* 96(2):137-149, 1994.
Contact: Dr. Tova Most, Hearing Impairment Program, School of Education and School of Communication Disorders, Tel-Aviv University, Israel

13. Hearing Aid Rejection: The Type III Individual. Piercy SK, Goldstein DP, *ASHA* 36(6/7):51-52, 1994.
Contact: Susan K. Piercy, Dept. Audiology and Speech Sciences, Purdue University, West Lafayette, IN 47907

14. Hearing Care in Nursing Homes Offers Daunting Challenges, Special Rewards. Bloom S, *Hear J* 47(8):13-20, 1994.
Contact: Sara Bloom, c/o The Hearing Journal, Williams & Wilkins, 428 East Preston St., Baltimore, MD 21202-3993

15. Influence of Aging on Active Cochlear Micromechanical Properties and on the Medial Olivocochlear System in Humans. Castor X, et al., *Hear Res* 77(1/2):1-8, 1994.
Contact: X. Castor, URA CNRS 1447, Physiologie sensorielle: 'Audition et voix', Hopital Edouard Herriot, 3 place d'Arsonval, F-69437 Lyon Cedex 03, France

COMMUNICATION AIDS—OTHER

16. Communication Disorders and Rehabilitation of Persons with Stroke. Rao PR, *Am Rehabil* 19(4):38-42, 1994.
Contact: Paul R. Rao, PhD, Speech-Language Pathology Service, Stroke Recovery Program, National Rehabilitation Hospital, Washington, DC 20010

COMMUNICATION AIDS—VISION

17. Cognitive Basis for Teaching Cane Travel. Mettler R, *J Visual Impairm Blindn* 88(4):338-348, 1994.
Contact: Richard Mettler, BA, Nebraska Dept. of Public Institutions, Division of Rehabilitation Services for the Visually Impaired, 4600 Valley Rd., Lincoln, NE 68510

18. Relation of Vision Loss to Depression in Older Veterans. Teitelbaum LM, et al., *J Visual Impairm Blindn* 88(3):253-257, 1994.
Contact: Lesley M. Teitelbaum, MA, Dept. of Clinical Psychology, Syracuse University, Syracuse, NY 13244

19. Research Abstracts on Blindness in India. Mokkapati J, *Bombay, India: NAB Louis Braille* Memorial Research Centre
Contact: Dr. Jayasree Mokkapati, Chief Officer (Researcher), NAB Louis Braille Memorial Research Center, 124 Cotton Depot, Cotton Green, Bombay-400 033 India 1993

20. Visual Cues for Enhancing Depth Perception. O'Donnell LM, Smith AJ, *J Visual Impairm Blindn* 88(3):258-266, 1994.
Contact: Lizabeth M. O'Donnell, MS, Institute for the Visually Impaired, Pennsylvania College of Optometry, 1200 Godfrey Ave., Philadelphia, PA 19141-3399

Listings #21 through #55 have been collected for this section by Gregory L. Goodrich, PhD, VA Western Blind Rehabilitation Center, Palo Alto, CA.

A Literature search was conducted using the keyword citation idex of *Low Vision—The References edited by Gregory L. Goodrich, PhD and Randy T. Jose, OD. Published by the Lighthouse Inc., New York, NY.*

Any questions concerning these entries should be directed to Dr. Goodrich at the VA Western Blind Rehabilitation Center, 3801 Miranda Ave., Palo Alto, CA 94304.

21. Adaption of the Bayley Scales of Infant Development for Visually Impaired Children. Smrkovsky M, et al., *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 121-128, 1994.

22. Assessment of Visually Impaired Infants. Hyvarinen L, *Ophthalmol Clin North Am* 7(2):219-25, 1994.

23. Cognitive and Metacognitive Interventions: Important Trends for Teachers of Students Who are Visually Impaired. Van Rueusen AK, Head DN *RE:view* 25(4):153-162, 1994.

24. Computer Programs for Visually Impaired Pre-School Children. Hammarlund J, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 338-344, 1994.

25. Congenital Optic Nerve Hypoplasia. Marsh-Tootle WL, Alexander LJ, *Opt Vis Sci* 71(3):174-181, 1994.

26. Creating Appropriate Toys for Children with Multiple Handicaps. Odess B, Margaliot S, *RE:view* 26(1):35-38, 1994.

27. Different Family—Having a Young Visually Impaired Child (= V.I.C.) and the Developments in a Family. Bonamie E, van Belle L, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 350-353, 1994.

28. Home Environment of Young Blind Children. Dote-Kwan J, Hughes M, *J Vis Impairm Blindn* 88(1):31-42, 1994.

29. How Should Computers be Used in the Visual Training of Young Visually Impaired Children? Bozic N, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 334-337, 1994.

30. Intelligence Profile of Children with Unequal Vision. Groenveld M, Jan JE, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 139-144, 1994.

31. ITVIC, an Intelligence Test for Visually Impaired Children. Boter RD, Hoekstra-Vrolijk SA, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 135-138, 1994.

32. Parents' Role Regarding Their Child with Low Vision. Liegner JT, *Ophthalmol Clin North Am* 7(2):227-230, 1994.

33. Pediatric Cataracts—Diagnosis and Treatment. Moore BD, *Opt Vis Sci* 71(3):168-73, 1994.

34. Place of Low Vision Devices for Students in Schools for Blind in Kenya and Uganda. Silver J, Gilbert C, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 400-403, 1994.

35. Playtherapy for Partially Sighted Children. van Gendt CJ, van Evert *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 369-371, 1994.

36. Practognostic Organization of Objects in the Immediate Environment—Applications with Young Visually and Multihandicapped Visually Impaired Children. Dewerchin L, *Low Vision: Research and New Developments in Rehabilitation.* Amsterdam: IOS Press, 316-319, 1994.

37. Quantitative Motor Diagnostics with Blind and Visually Impaired Children by Means of the Physical Coordination Test for Children. Reimer AM, Siemon-

sma-Boon M, Smits-Engelsman B, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press, 441-443, 1994.

38. Recruitment of Teachers of Visually Impaired Children: A Quest for Excellence. Alonso L, *J Vis Impairm Blindn* 88(4):117-125, 1994.

39. Rehabilitation of Visually Impaired Children in China. Sun B-C, Zhang S-y, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press, 506-509, 1994.

40. Relationship Between the Restricted Functional Near Visual Field of Partially Sighted Children and Reading Problems. Crisp RK, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press, 307-315, 1994.

41. Review of Embryology and Its Relation to Ocular Disease in the Pediatric Population. Trachimowicz RA, *Optom Vis Sci* 71(3):154-163, 1994.

42. Reynell-Zinkin Scales for Young Visually Handicapped Children: A Retrospective Study. Timmer-van de Vosse H, Hamers JMH, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press 129-134, 1994.

43. Role of Residential Schools for the Blind in 1990. McHahon E, *RE:view* 25(4):163-172, 1994.

44. Seeing Through the Isolation: A Study of First-Year Teachers of the Visually Impaired. Seitz JA, *J Vis Impairm Blindn* 88(4):299-309, 1994.

45. Spelling for the Visually Impaired Child. Arter C, Mason H, *Br J Vis Impairm* 12(1):18-21, 1994.

46. Survey of Children's First Understanding of Being Visually Impaired. Erin JN, Corn AL, *J Vis Impairm Blindn* 88(2):132-139, 1994.

47. Teaching Social Skills to Visually Impaired Students. Stockley J, *Br J Vis Impairm* 12(1):11-13, 1994.

48. Use of Dual Communication Boards with Students Who Are Deaf-Blind. Heller KW, et al., *J Vis Impairm Blindn* 88(4):368-376, 1994.

49. Use of the HOME Inventory with Families of Young Visually Impaired Children. Rock SL, et al., *J Vis Impairm Blindn* 88(2):140-151, 1994.

50. Vision Screening with the RDE Sterotest in Pediatric Populations. Schmidt PP, *Optom Vis Sci* 71(4):273-281, 1994.

51. Visual and Mentally-Multihandicapped Children and Adults: Training Programmes of Activities of Daily Living. van Dalden GW, Lopes C-H, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press, 456-460, 1994.

52. Visual Environmental Adaptation Problems of Partially Sighted Children. Kalloniatis M, Johnston AW, *J Vis Impairm Blindn* 88(3):234-243, 1994.

53. Visual Functioning: A Practical Approach for Teachers. Ekkens IE, *Low Vision: Research and New Developments in Rehabilitation*. Amsterdam: IOS Press, 24-27, 1994.

54. Visual Support to Multiply Visually Impaired Children in Learning Little Tasks. Andries B, *Low Vision: Research and New Developments in Rehabilitation*. IOS Press, 366-368, 1994.

55. Visually Impaired Children with Progressive, Terminal Neurodegenerative Disorders. Koehler W, Loftin M, *J Vis Impairm Blindn* 88(4):317-328, 1994.

FUNCTIONAL ELECTRICAL STIMULATION

56. Design and Clinical Application of a Double Helix Electrode for Functional Electrical Stimulation. Scheiner A, Polando G, Marsolais B, *IEEE Trans Biomed Eng* 41(5):425-431, 1994.
Contact: Avram Scheiner, Dept. of Veterans Affairs Medical Center, Motion Study Laboratory (151W), Cleveland, OH 44106

57. Effect of Participating in a Functional Electrical Stimulation Exercise Program on Affect in People with Spinal Cord Injuries. Bradley MB, *Arch Phys Med Rehabil* 75(6):676-679, 1994.
Contact: Mary Beth Bradley, MA, Blue Cross of Northeastern Pennsylvania, 70 North Main St., Wilkes-Barre, PA 18711

58. Implantable Stimulator for Selective Stimulation of the Common Peroneal Nerve: A Preliminary Report. Rozman J, et al., *J Med Eng Technol* 18(2):47-53, 1994.
Contact: Dr. J. Rozman, Jozef Stefan Institute, E-1, Biocybernetics Jamova 39, 61111 Ljubljana, Republic of Slovenia

59. Musculoskeletal Effects of an Electrical Stimulation Induced Cycling Programme in the Spinal Injured. Sloan KE, et al., *Paraplegia* 32(6):407-415, 1994.
Contact: K.E. Sloan, Dept. of Medical Physics, Royal Perth Hospital, Perth, Australia

See also 67

GAIT ANALYSIS

60. Adaptations During the Stance Phase of Gait for Simulated Flexion Contractures at the Knee. Cerny K, Perry J, Walker JM, *Orthopedics* 17(6):501-513, 1994.
Contact: Kay Cerny, PhD, Physical Therapy Dept., California State University, Long Beach, CA 90801

61. Comparison of Gait Characteristics in Young and Old Subjects. Ostrosky KM, et al., *Phys Ther* 74(7):637-646, 1994.
Contact: Karen M. Ostrosky, PT, Dept. of Physical Therapy, School of Health and Rehabilitation Sciences, University of Pittsburgh, 101 Pennsylvania Hall, Pittsburgh, PA 15261

62. Effects of Parallel Bars, Body Weight Support and Speed on the Modulation of the Locomotor Pattern of Spastic Paretic Gait. A Preliminary Communication. Visintin M, Barbeau H, *Paraplegia* 32(8):540-553, 1994.
Contact: Martha Visintin, MSc, Jewish Rehabilitation Hospital, Physiotherapy Dept., 3205 Place Alton Goldbloom, Chomedey, Laval, PQ H7V 1R2 Canada

63. Geometric Analysis of Muscle Mechanical Power with Applications to Human Gait. McFadyen BJ, *J Biomech* 27(9):1189-1193, 1994.
Contact: Bradford J. McFadyen, PhD, Professeur agrege, Dept. de Kinanthropologie, Universite du Quebec A Montreal, Case Postale 8888, Succ. A, Montreal, PQ H3C 3P8, Canada

64. Power Spectrum Analysis of the Rectified Electromyogram During Gait for Normals and Patients. Nielsen J, Arendt-Nielsen L, Pedotti A, *J Electromyograph Kinesiol* 4(2):105-115, 1994.
Contact: Jesper Nielsen, Laboratory for Motor Control, Dept. of Medical Informatics, Aalborg University, Fredrik Bajersvej 7D, 9220 Aalborg, Denmark

See also 68

GENERAL

65. Quantitative Approach to Modeling Mammalian Myelinated Nerve Fibers for Electrical Prosthesis Design. Frijns JHM, Mooij J, ten Kate JH, *IEEE Trans Biomed Eng* 41(6):556-566, 1994.
Contact: Johan H.M. Frijns, E.N.T. Dept., University Hospital Leiden, 2300 RC, Leiden, The Netherlands

66. Resource Guide—Products and Services for Incontinence. Jeter KF, *Spartinburg, SC: HIP* 1994:100 pages
Contact: Katherine F. Jeter, EdD, HIP, P.O. Box 8306, Spartanburg, SC 29305

See also 45

GERIATRICS

67. Adapting Older Persons' Homes to Promote Independence. Trickey F, et al., *Phys Occup Ther Geriatr* 12(1):1-14, 1994.
Contact: F. Trickey, MSc, Public Health Unit, Montreal General Hospital, 980 Guy St., Room 300A, Montreal, PQ H3H 2K3 Canada

68. Control of Body Centre of Mass Momentum During Sit-to-Stand Among Young and Elderly Adults. Pai Y-C, et al., *Gait & Posture* 2(2):109-116, 1994.
Contact: Yi-Chung Pai, PhD, 345 E. Superior St., Room 1323, Programs in Physical Therapy, Northwestern University Medical School, Chicago, IL 60611

69. Disability Following Hip Fracture. Craik RL, *Phys Ther* 74(5):387-398, 1994.
Contact: Rebecca L. Craik, PhD, PT, Program in Physical Therapy, Beaver College, Glenside, PA 19038-3295

70. Exercise Training and Nutritional Supplementation for Physical Frailty in Very Elderly People.

Fiatarone MA, et al., *N Engl J Med* 330(25):1769-1775 , 1994.
Contact: Maria A. Fiatarone, MD, Human Nutrition Research Center on Aging, 711 Washington St., Boston, MA 02111

71. Fractures in Older Persons. Thorngren K-G, *Disabil Rehabil* 16(3):119-126, 1994.
Contact: Prof. Karl-Goran Thorngren, MD, Dept. of Orthopedics, Lund University Hospital, S-221 85 Lund, Sweden

72. Predominance of Left-Sided Fractures of the Proximal Femur in a Geriatric Population. Rudolph V, *Orthopedics* 17(7):601-602, 1994.
Contact: Valerie Rudolph, BS, Physical Therapy Dept., New Rochelle Hospital Med. Ctr., 16 Guion Place, New Rochelle, NY 10801

73. Variant and Invariant Characteristics of the Sit-to-Stand Task in Healthy Elderly Adults. Vander Linden DW, Brunt D, McCulloch MU, *Arch Phys Med Rehabil* 75(6):653-660, 1994.
Contact: Darl W. Vander Linden, PhD, PT, Dept. of Physical Therapy, Eastern Washington University, West 407 Riverside, Suite 353, Spokane, WA 99201

See also 9,11,26,79

HEAD TRAUMA AND STROKE

74. Disability Outcomes Following Inpatient Rehabilitation for Stroke. Hamilton BB, Granger CV, *Phys Ther* 74(5):494-503, 1994.
Contact: Byron B. Hamilton, MD, PhD, Center for Functional Assessment Research, Dept. of Rehabilitation Medicine, School of Medicine and Biomedical Sciences, State University of New York at Buffalo, 232 Parker Hall, SUNY South Campus, Buffalo, NY 14214

75. Long Term Effects of Auditory Feedback Training on Relearned Symmetrical Body Weight Distribution in Stroke Patients: A Follow-Up Study. Engardt M, *Scand J Rehabil Med* 26(2):65-69, 1994.
Contact: Margareta Engardt, Dept. of Physical Therapy, Karolinska Institute, Huddinge, Sweden

76. Predicting Stroke Inpatient Rehabilitation Outcome Using a Classification Tree Approach. Falconer JA, et al., *Arch Phys Med Rehabil* 75(6):619-625, 1994.
Contact: Judith A. Falconer, PhD, OTR, Northwestern University, 345 East Superior St., Room 1323, Chicago, IL 60611

77. Stroke Disability. Duncan PW, *Phys Ther* 74(5):399-407, 1994.
Contact: Pamela W. Duncan, PhD, PT, Center for Health Policy Research and Education, Duke University, Erwin Sq., Ste 230, 2200 W. Main St., Durham, NC 27705

MUSCLES, LIGAMENTS, AND TENDONS

78. Comparison of Isometric and Load Moving Length—Tension Models of Two Bicompartmental Muscles. Vance TL, et al., *IEEE Trans Biomed Eng* 41(8):771-781, 1994.
Contact: Tedman L. Vance, Bioengineering Laboratory, Dept. of Orthopaedics, Louisana State University Medical Center, New Orleans, LA 70112

NEUROLOGICAL DISORDERS

79. Human Growth Hormone Effect on Serum IGF-I and Muscle Function in Poliomyelitis Survivors. Gupta KL, et al., *Arch Phys Med Rehabil* 75(8):889-894, 1994.
Contact: Kaup R. Shetty, MD, (111D), Zablocki VA Medical Center, 5000 W. National Ave., Milwaukee, WI 53295

80. Tardy Effect of Neurogenic Muscular Atrophy by Magnetic Stimulation. Chang C-W, Lien I-N, *Am J Phys Med Rehabil* 73(4):275-279, 1994.
Contact: Chein-Wei Chang, MD, Dept. of Physical Medicine & Rehabilitation, National Taiwan University Hospital, No. 1, Chang-Te St., Taipei 10016, Taiwan, Republic of China

ORTHOPEDICS

81. Bone Formation in Coralline Hydroxyapatite: Effects of Pore Size Studied in Rabbits. Kuhne J-H, et al., *Acta Orthop Scand* 65(3):246-252, 1994.
Contact: Dr. Jobst-Henner Kuhne, Dept. of Orthopedics, Klinikum Grobhadern, Marchioninistr. 15, W-81377 Munich, Germany

82. In Vitro Study of Proximal Femoral Allograft Strains in Revision Hip Arthroplasty. Ries MD, et al., *Med Eng Phys* 16(4):292-296, 1994.

Contact: Michael D. Ries, MD, Mary Imogene Bassett Hospital, Cooperstown, NY 13326

83. Lumbar Decompression to Treat Foot Drop After Hip Arthroplasty. Pritchett JW, *Clin Orthop* 303:173-177, 1994.
Contact: James W. Pritchett, MD, Orthopaedic Surgery, University of Washington, 1200 12th Ave. S., Seattle, WA 98144

84. Magnetic Resonance Imaging of the Lumbar Spine in People Without Back Pain. Jensen MC, et al., *N Engl J Med* 331(2):69-73, 1994.
Contact: Dr. Michael N. Brant-Zawadzki at Hoag Memorial Hospital, Dept. of Radiology, 301 Newport Blvd., Newport Beach, CA 92663

85. Neuronal Regulation of Fracture Healing: Effects of Sciatic Nerve Resection in Rat Tibia. Nordsletten L, et al., *Acta Orthop Scand* 65(3):299-304, 1994.
Contact: Dr. Lars Nordsletten, Institute of Surgical Research, Rikshospitalet (The National Hospital), N-0027 Oslo, Norway

86. Prospective, Double-Blind Trial of Electrical Capacitive Coupling in the Treatment of Non-Union of Long Bones. Scott G, King JB, *J Bone Joint Surg* 76A(6):820-826, 1994.
Contact: Garth Scott, FRCS, Bone and Joint Research Unit, The London Hospital Medical College, London E1 2AD, England

87. Removal of Bone Cement from the Femoral Canal Using an Acoustic Emission-Controlled Milling Device. Schmidt J, Nordmann K, *Med Biol Eng Comput* 32(3):258-260, 1994.
Contact: J. Schmidt, Oberarzt, Klinik und Poliklinik fur Orthopadie der Universitat zu Koln, Joseph-Stelzmann-Str. 9, D-5000 Koln 41, Germany

88. Talectomy for Osteoporotic and Neuropathic Feet—7 Cases Followed for 2-3 Years. Gunal I, *Acta Orthop Scand* 65(3):349-350, 1994.
Contact: Dr. Izqe Gunal, Sair Fuzuli Cad. Sokmen Sok, No. 13/3, 26010 Eskisehir, Turkey

89. Two-Year Outcome of Hydroxyapatite-Coated Prostheses: Two Femoral Prostheses Retrieved at Autopsy. Hardy DCR, et al., *Acta Orthop Scand* 65(3):253-257, 1994.
Contact: Dominique C.R. Hardy, Dept. of Orthopedics and Traumatology, University Hospital Saint-Pierre, Universite Libre de Bruxelles, Rue Haute, 322, B-1000 Brussells, Belgium

See also 78

PROSTHETICS

90. Below-Knee Amputee Gait in Stair Ambulation: A Comparison of Stride Characteristics Using Five Different Prosthetic Feet. Torburn L, et al., *Clin Orthop* 303:185-192, 1994.
Contact: Jacquelin Perry, MD, Rancho Los Amigos Medical Center, Pathokinesiology Laboratory, Bldg. 304, 7601 E. Imperial Hwy., Downey, CA 90242

91. Gait Analysis Study of an Energy-Storing Prosthetic Foot—A Preliminary Report. Goh JCH, et al., *Gait & Posture* 2(2):95-101, 1994.
Contact: Dr. J.C.H. Goh, Dept. of Orthopaedic Surgery, National University of Singapore, 5 Lower Kent Ridge Rd., Singapore 0511

92. Influence of Prosthesis Alignment on the Standing Balance of Below-Knee Amputees. Isakov E, et al., *Clin Biomech* 9(4):258-262, 1994.
Contact: Eli Isakov, MD, MSc, Biomechanics Laboratory, Loewenstein Rehabilitation Hospital, Ra'anana, PO Box 3, Israel 43100

93. Influence of Prosthetic Foot Design on Sound Limb Loading in Adults with Unilateral Below-Knee Amputations. Powers CM, et al., *Arch Phys Med Rehabil* 75(7):825-829, 1994.
Contact: Jacquelin Perry, MD, Rancho Los Amigos Medical Center, Pathokinesiology Laboratory, Bldg. 304, 7601 East Imperial Hwy., Downey, CA 90242

94. Method for Optimization of Above-Knee Prosthetic Shank-Foot Inertial Characteristics. Beck JC, Czerniecki J, *Gait Posture* 2(2):75-84, 1994.
Contact: Dr. J. Czerniecki, Human Motion Analysis Laboratory, VA Medical Center, Seattle, WA 98108

SPINAL CORD INJURY

95. Adjustment to Traumatic Spinal Cord Injury: A Longitunal Study of Self-Reported Quality of Life. Stensman R, *Paraplegia* 32(6):416-422, 1994.
Contact: R. Stensman, MD, Dept. of Rehabilitation Medicine, Uppsala University, Akademiska sjukhuset, S-751 85 Uppsala, Sweden

96. Augmentation of Transfers for a Quadriplegic Patient Using an Implanted FNS System: Case Report. Marsolais EB, et al., *Paraplegia* 32(8):573-579, 1994.
Contact: E.B. Marsolais, MD, PhD, Rehabilitation Medicine Service and Motion Study Laboratory, VA Medical Center, 10701 East Blvd., Cleveland, OH 44106

97. Bladder Function in the SCI Patient. Blaivas JG, *J Neurol Rehabil* 8(1):47-53, 1994.
Contact: Jerry G. Blaivas, MD, FACS, 400 East 56th St., New York, NY 10022

98. Correlation of Motor Control in the Supine Position and Assistive Device Used for Ambulation in Chronic Incomplete Spinal Cord-Injured Persons. Tang SFT, et al., *Am J Phys Med Rehabil* 73(4):268-274, 1994.
Contact: Simon F.T. Tang, MD, Dept. of Physical Medicine and Rehabilitation, 6550 Fannin, Suite 1421, Houston, TX 77030

99. Effect of Exercise Training on Pulmonary Function in Persons with Quadriplegia. Crane J, et al., *Paraplegia* 32(7):435-441, 1994.
Contact: L. Crane, MMSc, PT CCS, Division of Physical Therapy, University of Miami, Coral Gables, FL 33146

100. Factors Associated with the Quality of Life of Long-Term Spinal Cord Injured Persons. Clayton KS, Chubon RA, *Arch Phys Med Rehabil* 75(6):633-638, 1994.
Contact: Karen S. Clayton, PhD, Occupational Therapy Educational Department, Medical University of South Carolina, 171 Ashley Ave., Charleston, SC 29425

101. Isometric Shoulder Torque in Subjects with Spinal Cord Injury. Powers CM, et al., *Arch Phys Med Rehabil* 75(7):761-765, 1994.
Contact: Jacquelin Perry, MD, Rancho Los Amigos Medical Center, Pathokinesiology Laboratory, Bldg. 304, 7601 East Imperial Hwy., Downey, CA 90242

102. Life Satisfaction and Well-Being Measures in Ventilator Assisted Individuals with Traumatic Tetraplegia. Bach JR, Tilton MC, *Arch Phys Med Rehabil* 75(6):626-632, 1994.
Contact: John R. Bach, MD, Dept. of Physical Medicine and Rehabilitation, University Hospital B-239, 150 Bergen St., Newark, NJ 07939

103. Management of the Altered Physiology of the SCI Patient. Young RR, *J Neurol Rehabil* 8(1):25-32, 1994.
Contact: Robert R. Young, MD, Dept. of Neurology, University of California at Irvine, 105 Irvine Hall, Irvine, CA 92717-4275

104. Neurorehabilitation of Spinal Cord Injury. Young W, *J Neurol Rehabil* 8(1):3-9, 1994.
Contact: Wise Young, PhD, MD, Dept. of Neurosurgery, New York University Medical Center, 550 First Ave., New York, NY 10016

105. Prognostic Value of Cortical Magnetic Stimulation in Spinal Cord Injury. Clarke CE, et al., *Paraplegia* 32(8):554-560, 1994.
Contact: C.E. Clarke, BSc, MD, MRCP, Dept. of Neurology, Hull Royal Infirmary, Anlaby Rd., Hull HU3 2JZ, England

106. Regenerative Potential of the Spinal Cord. Selzer ME, *J Neurol Rehabil* 8(1):11-16, 1994.
Contact: Michael E. Selzer, MD, PhD, Dept. of Neurology, Hospital of the University of Pennsylvania, 3400 Spruce St., Philadelphia, PA 19104-4283

107. Reliability of Heart Rate Responses to Non-Steady-State Activities of Daily Living in Men with Spinal Cord Injuries. Janssen TWJ, et al., *Scand J Rehabil Med* 26(2):71-78, 1994.
Contact: Thomas W.J. Janssen, MSc, Faculty of Human Movement Sciences, Vrije University, Amsterdam, The Netherlands

108. Stabilization of Thoracic Spine Fractures Resulting in Complete Paraplegia—A Long-Term Retrospective Analysis. Place HM, et al., *Spine* 19(15):1726-1730, 1994.
Contact: Maj. Howard M. Place, MD, Orthopedic Service, Fitzsimons Army Medical Center, Aurora, CO 80045-5001

109. Voiding Alert System: A New Application in the Treatment of Incontinence. Murphy KP, Kliever EM, Moore MJ, *Arch Phys Med Rehabil* 75(8):924-927, 1994.
Contact: Kevin P. Murphy, MD, Dept. of Physical Medicine and Rehabilitation, Duluth Clinic, Limited, 400 East Third St., Duluth, MN 55805

See also 28,82

SURGERY

110. Complications of Spinal Surgery in Acute Spinal Cord Injury. Carvell JE, Grundy DJ, *Paraplegia* 32(6):389-395, 1994.
Contact: J.E. Carvell, MMSc FRCS, Salisbury District Hospital, Salisbury SP2 8BJ, England

111. Operative and Nonoperative Management of Spinal Cord Injury. A Review. Donovan WH, *Paraplegia* 32(6):375-388, 1994.
Contact: W.H. Donovan, MD, Dept. of Physical Medicine and Rehabilitation, University of Texas Health Science Center-Houston, Houston TX 77030

WHEELCHAIRS and POWERED VEHICLES

112. Physical Strain in Daily Life of Wheelchair Users with Spinal Cord Injuries. Janssen TWJ, et al., *Med Sci Sports Exerc* 26(6):661-670, 1994.
Contact: Thomas W.J. Janssen, Faculty of Human Movement Sciences, Vrije Universiteit Amsterdam, 1081 BT Amsterdam, The Netherlands

WOUNDS and ULCERS

113. Pressure Ulcers: A Review. Yarkony GM, *Arch Phys Med Rehabil* 75(8):908-917, 1994.
Contact: Gary M. Yarkony, MD, Schwab Rehabilitation Hospital, 1401 S. California Blvd., Chicago, IL 60608

114. Understanding the Pressure Ulcer Problem. Feedar JA, *Top Geriatr Rehabil* 9(4):1-7, 1994.
Contact: Jeffrey A. Feedar, BS, PT, Wound Care Assoc., Williams Bay, WI 53191

115. Wound Evaluation and Treatment Planning. Feedar JA, *Top Geriatr Rehabil* 9(4):35-42, 1994.
Contact: Jeffrey A. Feedar, BS, PT, Wound Care Assoc., Williams Bay, WI 53191

116. Wound Healing Model and Clinical Applications. Feedar JA, *Top Geriatr Rehabil* 9(4):8-16, 1994.
Contact: Jeffrey A. Feedar, BS, PT, Wound Care Associates, Williams Bay, WI 53191

Periodicals reviewed for PUBLICATIONS OF INTEREST

Acta Orthopaedica Scandinavica
Advances in Orthopaedic Surgery
American Journal of Occupational Therapy
American Journal of Physical Medicine and Rehabilitation
American Journal of Sports Medicine
American Rehabilitation
Annals of Biomedical Engineering
Archives of Physical Medicine and Rehabilitation
ASHA (American Speech and Hearing Association)
Assistive Technology
Biomaterials, Artificial Cells, and Immobilization Biotechnology
Biomedical Instrumentation & Technology
Canadian Journal of Rehabilitation
Clinical Biomechanics
Clinical Kinesiology
Clinical Orthopaedics and Related Research
Clinical Rehabilitation
CRC Critical Reviews in Biomedical Engineering
DAV Magazine (Disabled American Veterans)
Disability and Rehabilitation
Electromyography and Clinical Neurophysiology
Engineering in Medicine and Biology Magazine
Ergonomics
Gait & Posture
Hearing Journal
Hearing Rehabilitation Quarterly
Hearing Research
Human Factors: The Journal of the Human Factors Society
IEEE Engineering in Medicine and Biology Magazine
IEEE Transactions on Biomedical Engineering
IEEE Transactions in Systems, Man and Cybernetics
IEEE Transactions on Rehabilitation Engineering
International Journal of Rehabilitation Research
JAMA
Journal of Acoustical Society of America
Journal of Applied Biomaterials

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Journal of Biomechanics
Journal of Biomedical Engineering
Journal of Biomedical Materials Research
Journal of Bone and Joint Surgery—American Ed.
Journal of Bone and Joint Surgery—British Ed.
Journal of Clinical Engineering
Journal of Electromyography and Kinesiology
Journal of Head Trauma and Rehabilitation
Journal of Medical Engineering and Technology
Journal of Neurologic Rehabilitation
Journal of Orthopaedic and Sports Physical Therapy
Journal of Orthopaedic Research
Journal of Prosthetics and Orthotics
Journal of Rehabilitation
Journal of Speech and Hearing Research
Journal of Trauma
Journal of Vision Rehabilitation
Journal of Visual Impairment and Blindness
The Lancet
Medical and Biological Engineering and Computing
Medical Psychotherapy Yearbook
Medicine & Science in Sports and Exercise
Military Medicine
New England Journal of Medicine
The Occupational Therapy Journal of Research
Orthopaedic Review
Orthopedic Clinics of North America
Orthopedics
Paraplegia
Paraplegia News
Physical and Occupational Therapy in Geriatrics
Physical Medicine and Rehabilitation
Physical Therapy
Physiotherapy
Proceedings of the Institution of Mechanical Engineers—Part H: Journal of Engineering in Medicine
Prosthetics and Orthotics International
Rehab Management
Rehabilitation Digest
Scandinavian Journal of Rehabilitation Medicine
Science
Spine
Sports 'N Spokes
Techniques in Orthopaedics
Topics in Geriatric Rehabilitation
VA Practitioner
Vanguard
Volta Review

CALENDAR OF EVENTS

NOTE: An asterisk at the end of a citation indicates a new entry to the calendar.

1994

December 7-10, 1994
8th International Conference on Biomedical Engineering, Singapore
Contact: The Secretary, 8th ICBME 1994, 336 Smith Street, #06-302, New Bridge Centre, Singapore, 0105; Tel: (65) 227-9811; Fax: (65) 227-0257

1995

February, 1995
Conference on Orientation and Navigation Systems for Blind Persons, London, England
Contact: Dr. John Gill, Royal National Institute for the Blind, 224 Great Portland Street, London W1N 6AA, England; Fax: + 44 71 388 7747

February 16-18, 1995
11th International Seating Symposium, Pittsburgh, Pennsylvania
Contact: Elaine Trefler or Jill Bebout, University of Pittsburgh Medical Center, Department of Conference Management, Nese-Barkan Bldg., Suite 511, Pittsburgh, PA 15213-2593; Tel: 412-647-8218; Fax: 412-647-8222

February 16-21, 1995
American Academy of Orthopedic Surgeons Annual Convention, (AAOS) Orlando, Florida
Contact: American Academy of Orthopedic Surgeons, 222 S. Prospect Avenue, Park Ridge, IL 60068; Tel: 708-823-7186

February 16-21, 1995
American Association for the Advancement of Science (AAAS), Atlanta, Georgia
Contact: AAAS, 1333 H. Street, NW, Washington, DC 20005; Tel: 202-326-6400

March, 1995
Vth International Conference for Surgical Rehabilitation of Upper Limb in Tetraplegia (Quadraplegia), Melbourne, Australia
Contact: Gerard Sormann, MBBS, FRACS, The Medical Centre, 517 St. Kilda Road, Melbourne 3004, Australia; Tel: 866-8668; Fax: 867-8637

March 21-25, 1995
American Academy of Orthotists and Prosthetists Annual Meeting (AAOP), New Orleans, Louisiana
Contact: Annette Suriani, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7116

March 27-31, 1995
12th World Congress of the International Federation of Physical Medicine and Rehabilitation (IFPMR), Sydney, Australia
Contact: Dianna Crebbin Conferences, PO Box 629, Willoughby NSW 2068, Australia; Tel: + 61 (02) 417-8525; Fax: + 61 (02) 417-8513

April 2-7, 1995
8th World Congress of the International Society for Prosthetics and Orthotics (ISPO), Melbourne, Australia
Contact: Congress Secretariat, 84 Queensbridge Street, South Melbourne, Victoria, Australia 3205; Tel: + 61 + 3-682-0244; Fax: + 61 + 3-682-0288

April 8-9, 1995
14th Southern Biomedical Engineering Conference, Shreveport, Louisana
Contact: Dr. Debi P. Mukherjee; Tel: 318-674-6187; Fax: 318-674-6186

April 9-11, 1995
Japanese Orthopaedic Association: 68th Annual Meeting, Yokohama City, Japan
Contact: Dr. T. Kurokawa, President, Department of Orthopaedic Surgery, Faculty of Medicine, The University of Tokyo, 7-3-1 Hongo, Bunkyo-ku, 113, Japan

April 25-28, 1995
INTERHOSPITAL '95, Hannover, Germany
Contact: Donna Peterson Hyland; Tel: 609-987-1202; Fax: 609-987-0092*

May 1-3, 1995
American Spinal Cord Injury Association Annual Meeting (ASIA), Orlando, Florida
Contact: ASIA; Tel: 404-355-9772

May 4-6, 1995
2nd International Workshop on Implantable Telemetry, Measurement of Mechanical Parameters in Humans, Berlin, Germany
Contact: Dr.-Ing. Georg Bergmann, Oskar-Helene-Heim, Biomechanics-Laboratory, Orthopaedic Hospital of the Free University, Clayallee 229, D—14195 Berlin, Germany; Tel: Int. + 49 30 81004-273; Fax: Int. + 49 30 81004-207*

May 27-31, 1995
1st International Rehabilitation Medicine Congress, Istanbul, Turkey
Contact: Dr. Onder Kayhan, Congress Secretariat, P.O. Box, Kosuyolo 81121, Istanbul, Turkey; Tel: 216-326-3443; Fax: 216-326-3444*

May 28-June 1, 1995
5th European Congress on Research in Rehabilitation, Helsinki, Finland
Contact: Professor Simon Miller, Division of Clinical Neuroscience, The Medical School, The University, Newcastle upon Tyne, NE2 4HH, United Kingdom; Tel: + 44 91 222 6617; Fax: + 44 91 222 8803*

June 9-14, 1995
RESNA International Conference Vancouver, BC
Contact: RESNA; Tel: 703-524-6686

June 22-25, 1995
Annual Meeting of the American Congress of Rehabilitation Medicine, Arlington, Virginia
Contact: American Congress of Rehabilitation Medicine, 5700 Old Orchard Road, First Floor, Skokie, IL 60077-1057; Tel: 708-966-0095; Fax: 708-966-9418*

July 9-16, 1995
4th World Congress of Neuroscience, Kyoto, Japan
Contact: Host Organizer, Secretariat, 4th World Congress of Neuroscience, c/o International Communications, Inc., Kasho Bldg., 2-14-9, Nihonbashi, Chuo-ku, Tokyo 103, Japan; Tel: 03-3272-7981; Fax: 03-3273-2445

July 16-19, 1995
7th International Conference on Mobility and Transport for Elderly and Disabled People, Reading, England
Contact: 7th International Conference Secretariat, Disability Unit, Department of Transport, Room S10/21, 2 Marsham Street, London SW1P 3EB, England

July 22-27, 1995
3rd International Neurotrauma Symposium, Toronto, Canada
Contact: Conference Secretariat c/o: Congress Canada, 191 Niagara Street, Toronto, Ontario, Canada, M5V 1C9; Tel: 416-860-1772; Fax: 416-860-0380

September 5-8, 1995
Second Leeds European Rehabilitation Conference Neurological Rehabilitation: New Initiatives in Treatment & Measuring Outcome, Leeds, England UK
Contact: Mrs. Carol Would, Conference Secretary, Department of Continuing Professional Education, Continuing Education Building, Springfield Mount, Leeds LS2 9NG; Tel: (0532) 333232; Fax: (0532) 333240

September 8-10, 1995
4th Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Oslo, Norway
Contact: Congress Secretariat, 4th Scientific Meeting of SMSOP, c/o Sunnaas Hospital, N-1450 Nesoddtangen, Norway; Tel: + 47 66 96 90 00; Fax: + 47 66 91 25 76

September 11-13, 1995
First Biennial Conference, Advancing Human Communication: An Interdisciplinary Forum on Hearing Aid Research and Development, Bethesda, Maryland
Contact: NIDCD; Tel: 301-496-7243; TDD 301-402-0252

September 11-19, 1995
10th Asia Pacific Regional Conference of Rehabilitation International, Indonesia
Contact: Secretariat, 10th ASPARERI, H.Hang, jebat II-2 Blok F IV, Kebayoran Baru, Jakarta 12120, Indonesia; Tel: + 62 21 717 366*

September 19-23, 1995
American Orthotic and Prosthetic Association, National Assembly (AOPA), San Antonio, Texas
Contact: Annette Suriani, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7116

September 20-23, 1995
17th Annual International Conference of the IEEE Engineering in Medicine and Biology Society and 21st Canadian Medical and Biological Engineering Conference, Montreal, Canada
Contact: Robert E. Kearney, PhD, Eng, Department of Biomedical Engineering, McGill University, 3775 Univer-

sity Street, Montreal, Quebec, Canada H3A 2B4; Tel: 514-398-6737; Fax: 514-398-7461; E-Mail: rob@neuron.biomed.mcgill.ca

November 17-20, 1995
American Speech-Language-Hearing Association (ASHA), Annual Convention, Cincinnati, Ohio
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; Tel: (301) 897-5700

November 17-22, 1995
American Academy of Physical Medicine & Rehabilitation (AAPM&R), Orlando, Florida
Contact: AAPM&R; Tel: 312-922-9366

1996

February 22-27, 1996
American Academy of Orthopedic Surgeons Annual Convention(AAOS), Atlanta, Georgia
Contact: AAOS, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7114

April 22-May 5, 1996
18th World Congress of Rehabilitation International Equality Through Participation—2000 and Beyond, Auckland, New Zealand
Contact: Mrs. Bice Awan, Accident Rehabilitation & Compensation, Insurance Corporation, PO Box 242, Wellington, New Zealand; Tel: + 64 4 4738 775*

May 12-16, 1996
The First Mediterranean Congress on Physical Medicine and Rehabilitation, Herzlia, Israel
Contact: Dr. Haim Ring, c/o Ortra Ltd., PO Box 50432, Tel Aviv 61500, Israel; Tel: 972-3-664825; Fax: 972-3-660952

August 7-10, 1996
7th International ISAAC Conference on Augmentative and Alternative Communication, Vancouver, BC, Canada
Contact: ISAAC, PO Box 1762, Station R., Toronto, Ontario, Canada; Tel: + 1 416 737 9308*

August 16-27, 1996
1996 Atlanta Paralympic Games, Atlanta, Georgia
Contact: Tel: 404-588-1996*

1997

February 13-18, 1997
American Academy of Orthopedic Surgeons Annual Convention (AAOS), San Francisco, California
Contact: AAOS, 1650 King Street, Suite 500, Alexandria, VA 22314; Tel: 703-836-7114

August 31-September 5, 1997
8th World Congress of the International Rehabilitation Medicine Association IRMA, Kyoto, Japan
Contact: Japan Convention Services, Inc., Nippon Press Center Bldg., 2-1, 2-chome, Uchisaiwai-cho, Chiyoda-ku, Tokyo 100, Japan*

LETTERS TO THE EDITOR

Re: **Three-dimensional computer model of the human buttocks, in vivo by Beth A. Todd, PhD and John G. Thacker, PhD, Vol 31, No. 2, pp. 111-9, 1994.**

To the Editor:

Characterizing the internal state of loaded human tissue *in vivo* is a daunting task as we know firsthand from our modeling and measurement of matrix pressure and fluid pressure and flow in synovial joints (e.g., "Cartilage Pressures in the Human Hip Joint," Macirowski T., Tepic S. and Mann R.W., *Journal of Biomechanical Engineering*, February 1994, Vol. 116, pp. 10-18). Your effort to characterize the human buttock under load, especially given the serious and poorly understood etiology of the scourge of bed and wheelchair bound paralyzed persons (i.e., decubitus ulcers) is to be commended.

You report in your paper and "Clinical Relevance" summary that "for the male subject, stress was 3.5 to 4.5 times greater in internal tissues than at the buttock-cushion surface. For the female subject, stress in internal tissues was 11 to 13.5 times greater." These finite-element model estimates are presented in Table 4.

I am intrigued that gender can produce a factor-of-three change in tissue pressure, even given the morphology and weight differences between your male subject and your female subject. Somehow it doesn't seem possible that the biology and physiology of tissue can be that different between man and woman. Perhaps you have thought about this, or can attribute the difference to aspects of the model.

Your "Summary" states that "Computational results were verified experimentally with magnetic resonance imaging and interface pressure measurements." In Table 3 you compare the model estimates of stress with the Oxford pressure monitor measurements and note that the "computational principal stresses are approximately 50 percent different from the experimentally determined pressures." My reading of your Male-Supine comparison has the measurement 37 percent *higher* than the model estimate, while in the comparison Female-Supine the measurement is 40 percent *lower* than the computational result. I appreciate that, as you say, "As more information . . . is introduced . . . the results will improve," but I am perplexed by the sign difference between the two comparisons.

Clearly this important area, understanding how contact pressure influences ulcer development, requires more work, as you say. My overarching concern at this stage is the possible misreading of your research in the "Summary of Scientific/Technical Papers in this Issue" which is headed "Clinical Relevance for the Veteran." A lay person unfamiliar with finite-element analysis, but concerned with the seating problems of the insensate paralyzed person, might assume we are closer to that understanding than your work, and that of others, warrants.

Sincerely yours,
Robert W. Mann
Whitaker Professor Emeritus
Biomedical Engineering
Massachusetts Institute of Technology

The Author Replies:

We appreciate the interest and comments from Professor Mann. As he has indicated, modeling internal tissues is a difficult task, and he has cited evidence that alludes to areas which beckon researches for further study: Why is the difference between skin surface and ischial tuberosity pressures so much greater for the female than the male? What would cause the sign differences between experimental and computational results for the two subjects?

To begin with his second point, changes in the boundary conditions that are applied to the models made a significant difference. The boundary conditions were chosen so that they would be the same for both models while matching the experimental data as closely as possible. While the constraints on the cushion, symmetry condition, and elastic foundation were all reasonable assumptions, computer memory limitations necessitated the inferior and superior surfaces of the soft tissue in the model to remain free. The advent of the PowerPC allows more flexibility in the model.

Regarding the first point, it is believed that morphology was the primary reason for the factor-of-three differences between skin surface and ischial tuberosity stresses. However that is only a belief. With such a small sample size (ONE male and ONE female), there are many variables which could affect the stresses that may not be related to gender at all.

Currently, funding is being sought to continue this study with a large population. With the rapid advances

occurring in computer hardware, a more detailed model can be solved on a desktop platform. This will also allow the exploration of the nonlinear aspects of the tissues.

Finally, Professor Mann concludes with concern that a nontechnical reader may be mislead by the general comments made in the "Summary of Scientific/Technical Papers in this Issue." We regret the emphasis that we placed on the word summary when writing this section. It is important for all authors to remember to read their work from the layperson's point of view, and we should have taken more care in our writing.

While the work presented in this paper brings us closer to an understanding of the transmission of forces through the human buttocks, it is only a very tiny step. There continues to be much work to be done to reach a full understanding of this problem and the end of seating problems for the disabled population.

Beth A. Todd, PhD
Assistant Professor
College of Engineering
University of Alabama

*U.S. G.P.O.: 1994-301-908:00001

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